Maria Sklodowska-Curie
National Research
Institute of Oncology

TPS algorithms
part II – real situation

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Approximations

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Approximations

Two scenarios: transient charged equilibrium
exist
does not exist

Transient charged equilibrium exist

- to know fluence is enough
  - it happens
  - if distance from the point of interest located in tissue A to the nearest point in tissue B is larger than electrons range,
    - (Air - soft tissue interface, lung – soft tissue interface, etc.)
  - If the distance from the point of interest to the region where the fluence is really different is larger than electrons range
    - penumbra problem

What is the range of electrons generated in 6 MV photon beam in water?

- in water
- in lung of density of 0.2 g/cm3

Radiological depth

Radiological distance

Distance

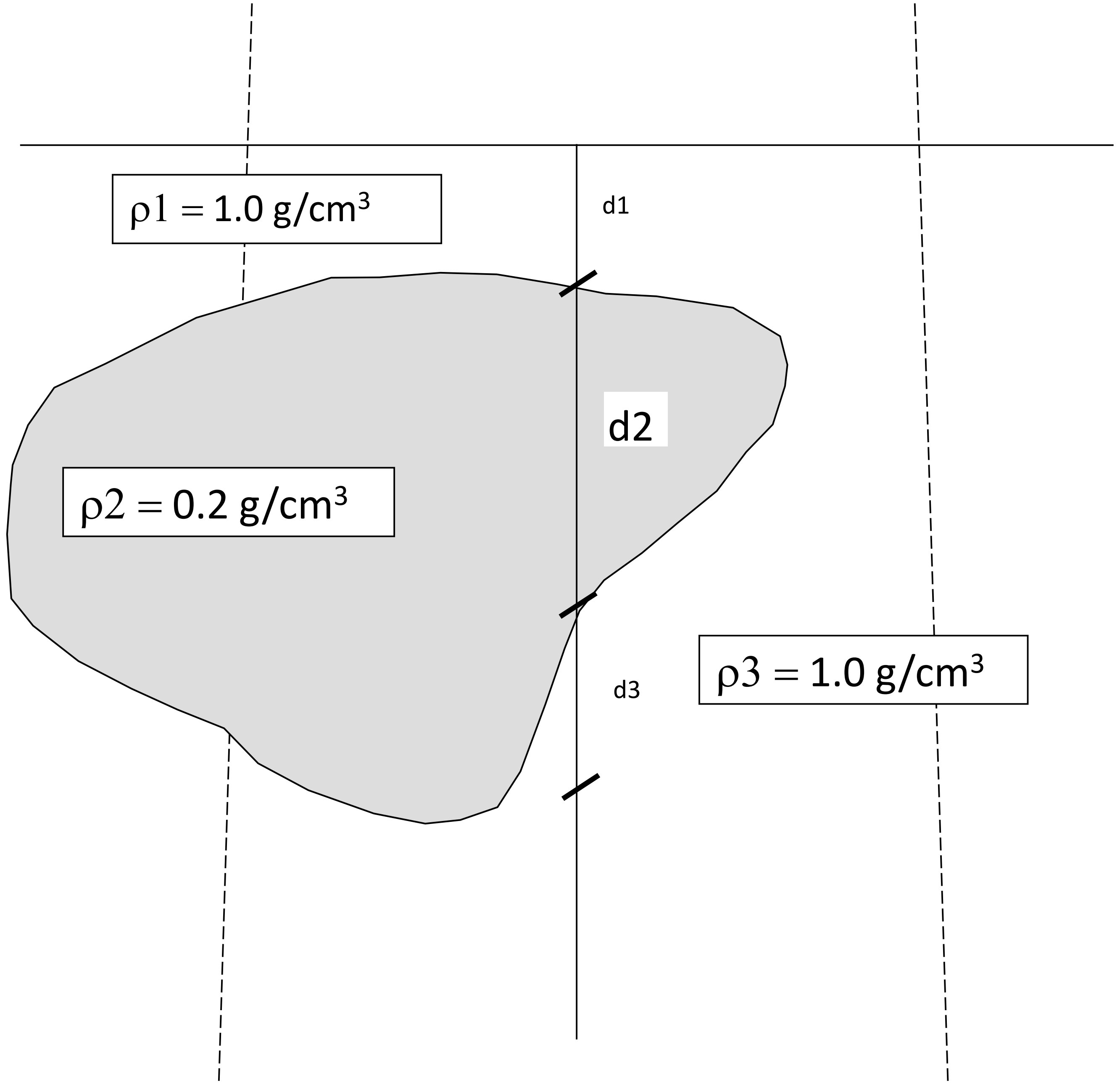
scaled with density

this methodology is applied either for photons (fluence) or for electron tracs (dose)

CPE inhomogeneous aborbent

correction factor for inhomogeneities

- The radiological distance
  - the thickness of the material of unit denisity (water) which attenuates in the same way as the heterogeneous absorbent
    - physical distance to the point remains the same



approximation

$$d_{rad} = \rho 1 \cdot d1 + \rho 2 \cdot d2 + \rho 3 \cdot d3$$

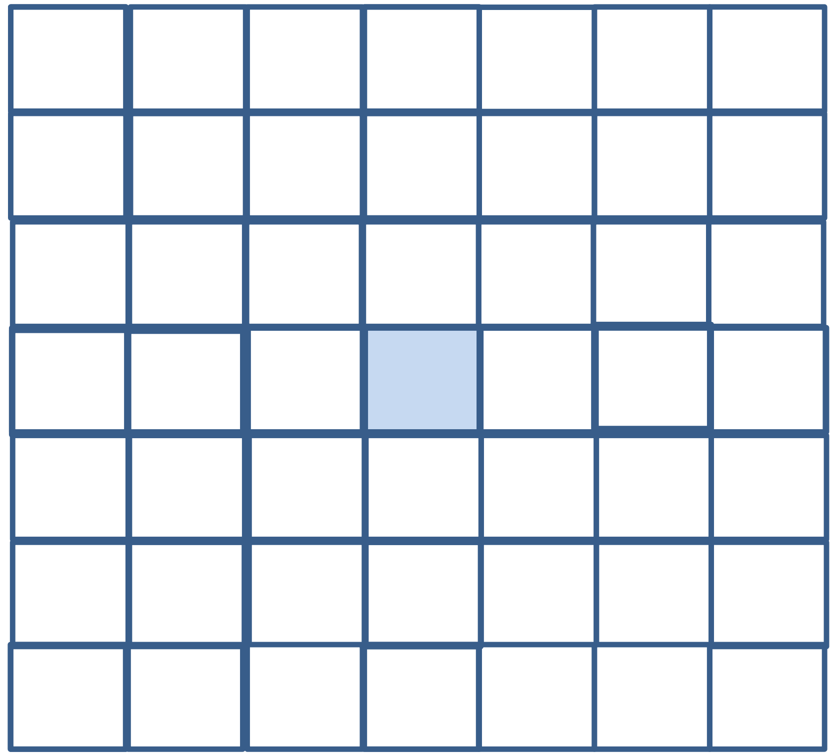
Transient Charged Equilibrium exists

Dose distribution calculation is (quite) simple

- dose is proportional to fluence
  - inverse square factor and attenuation
- **energy transfered is given by**

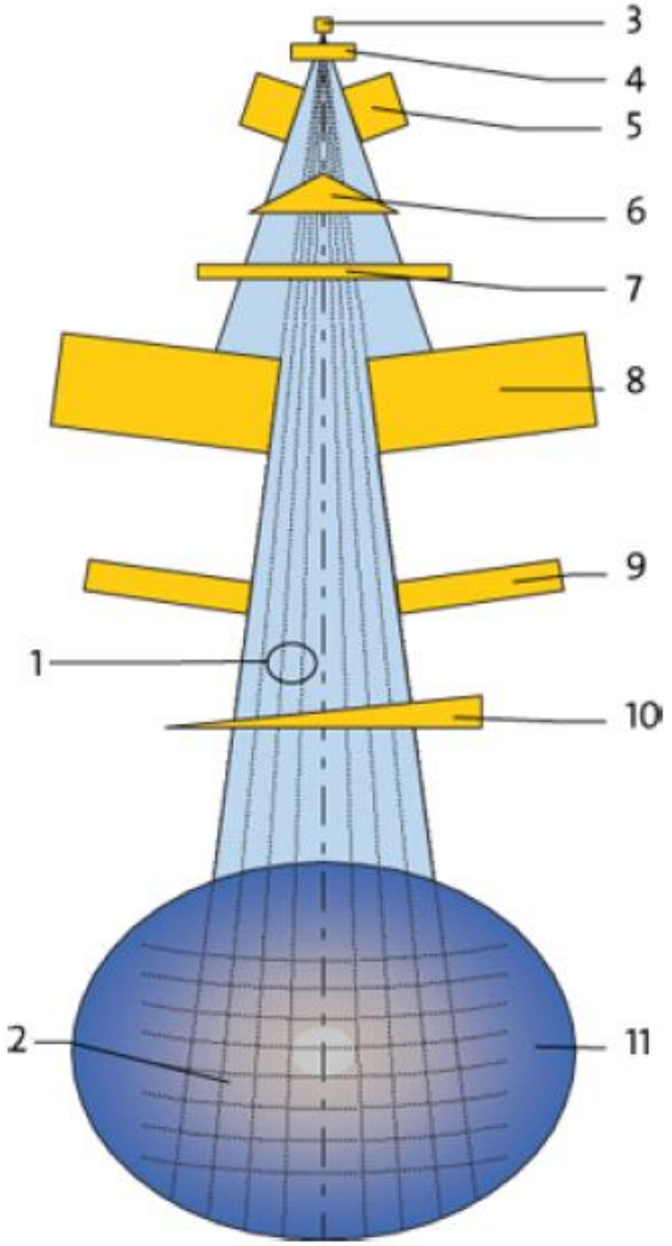
$$T(d) = TERMA_{h\nu}(d) = \Phi^{F}_{air} \cdot \frac{F^2}{(F+f)^2} \cdot e^{-\mu_{h\nu} d} \cdot h\nu \cdot \left(\frac{\mu_{h\nu}}{\rho}\right)$$

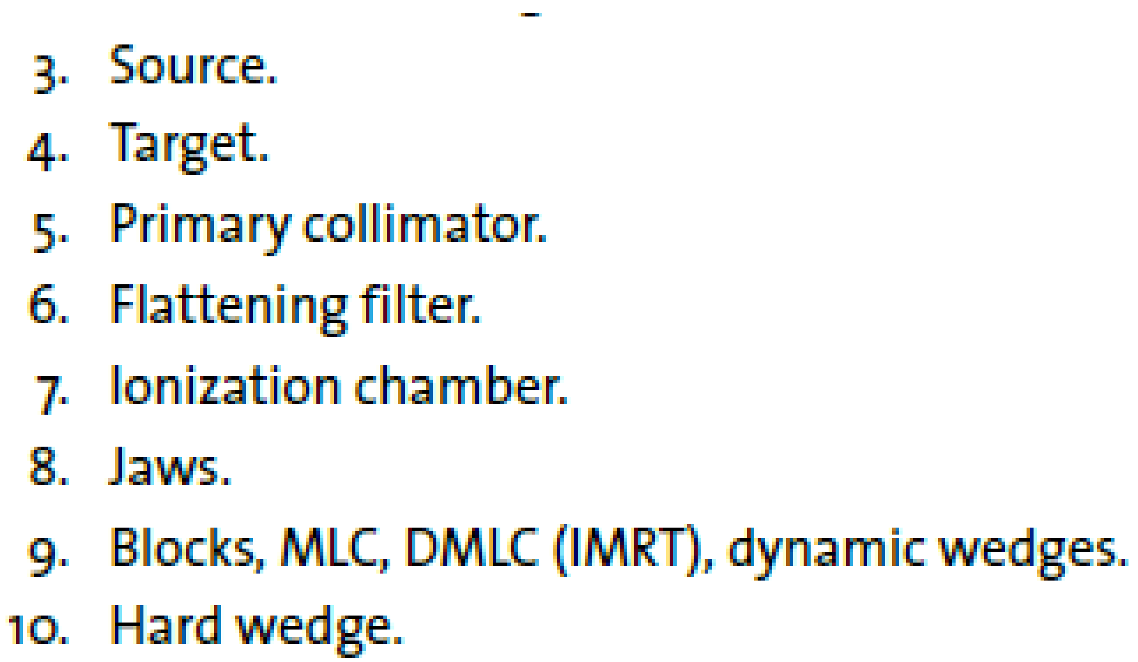
$$T(d) = \int_{spectrum} \frac{d\Phi^{F}_{air}}{dh\nu} \cdot \frac{F^2}{(F+f)^2} \cdot e^{-\mu_{h\nu} d} \cdot h\nu \cdot \left(\frac{\mu_{h\nu}}{\rho}\right) dh\nu$$



Fluence modeling

Eclipse




3. Source.
4. Target.
5. Primary collimator.
6. Flattening filter.
7. Ionization chamber.
8. Jaws.
9. Blocks, MLC, DMLC (IMRT), dynamic wedges.
10. Hard wedge.

User Guide Eclipse

Approximation

Photon fluence is decomposed into

- primary photon source,
- second photon source,
- electron contamination source,
- photons scattered from the hard wedge (wedge scatter source)

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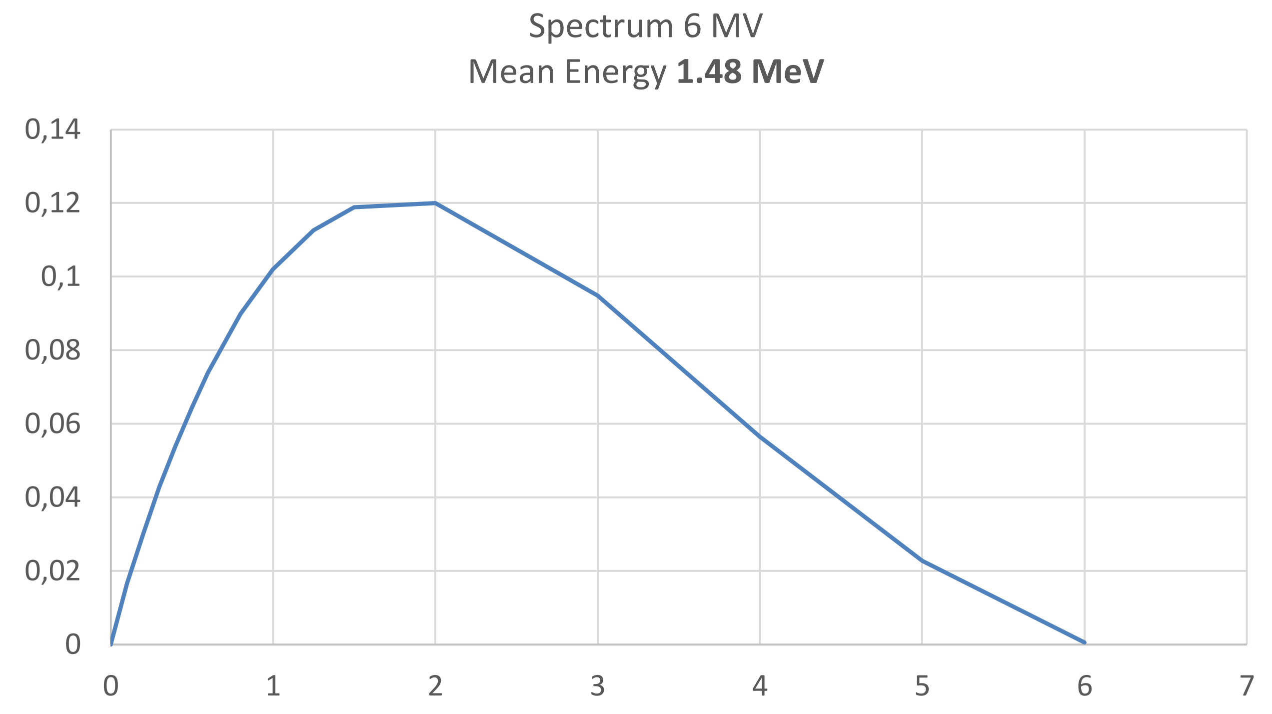
Approximations energy spectrum

Calculating the integral is a task beyond the capabilities of current computers in a reasonable amount of time

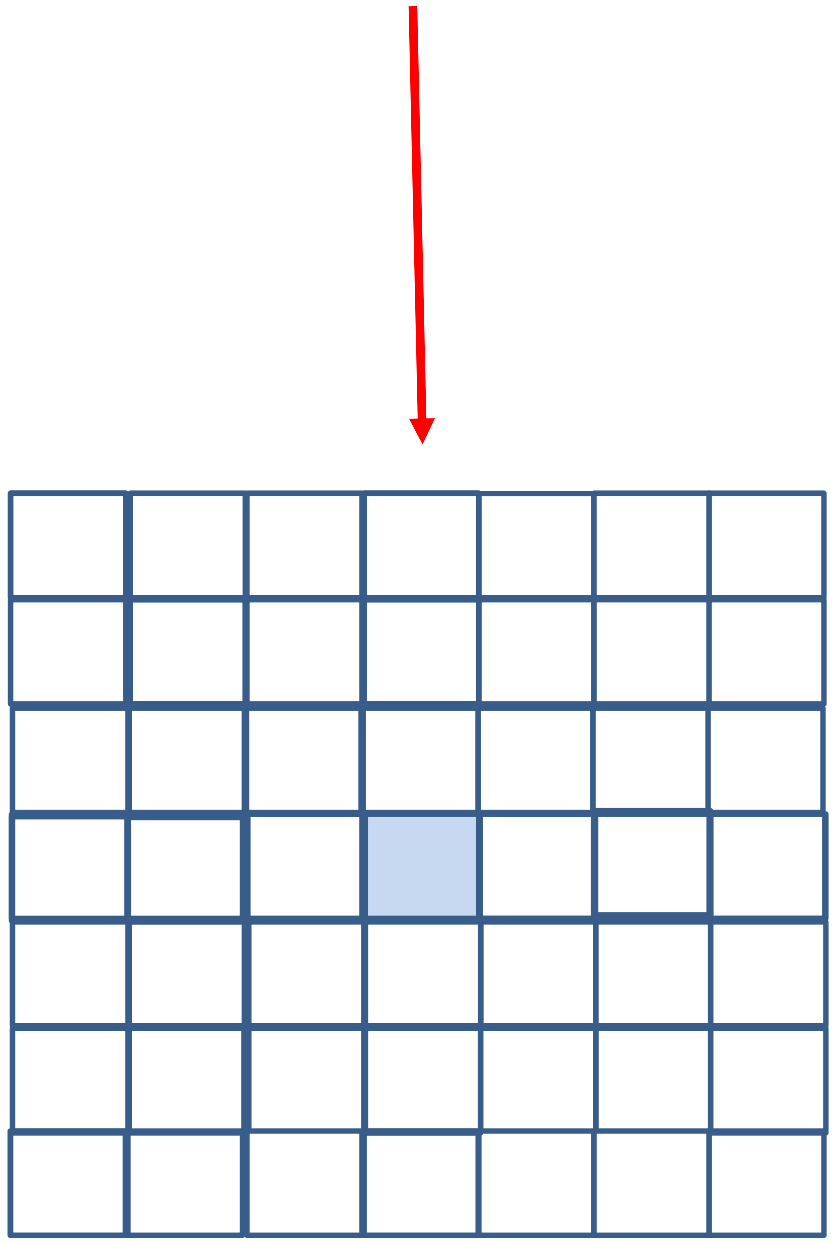
Different approximations are used

- monoenergetic beam is used (kernel)
- integral is not calculated over full space
- Kernels are calculated for water only
    - distance scaling is applied



Eclipse
mean energy as a function of the radius from the central axis

Kernels (A)
dose distribution from „small voxel”

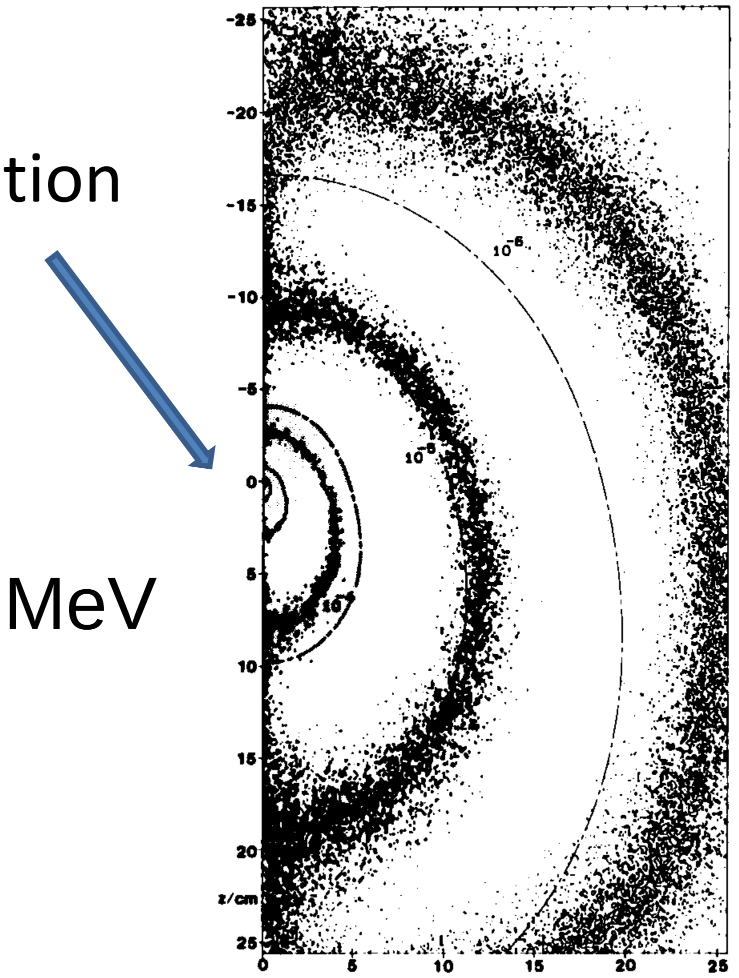


$$A_{hv}(\overline{r}-\overline{r'};hv)$$

scattered

interaction

X 0,4 MeV



Mohan, Med.Phys, 1985, 12, 592 – 597.

Kernels

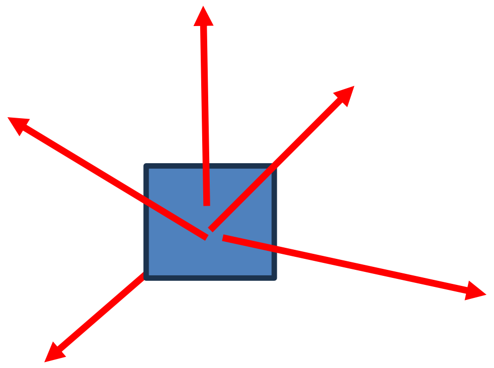
Kernels can be analytically described with high precision by

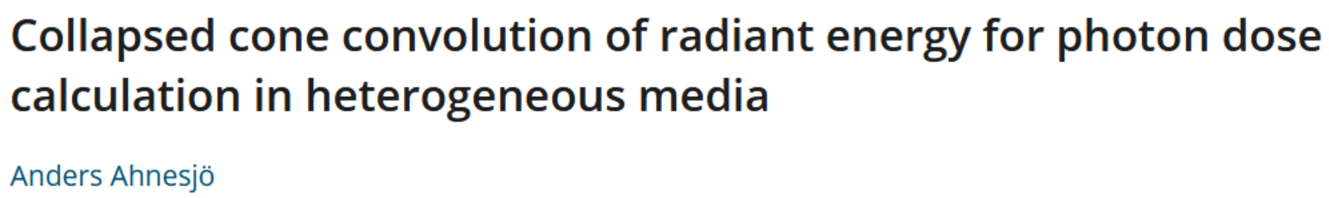
$\{A \exp(-ar)+B \exp(-br)\}/r^2$, where A, a, B, and b

depend on the angle with respect to the impinging photons and the

accelerating potential, and r is the radial distance.

radial distance is scaled with density

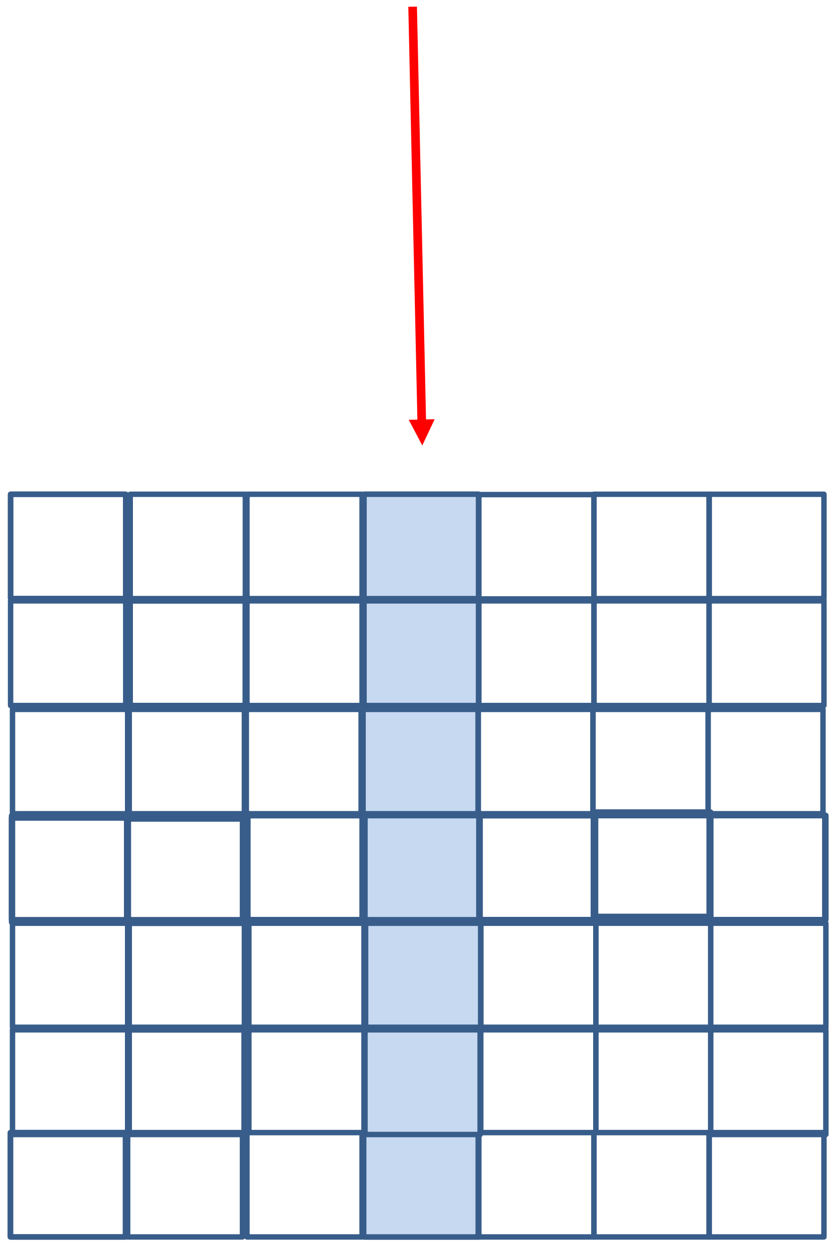




Collapsed cone convolution of radiant energy for photon dose calculation in heterogeneous media

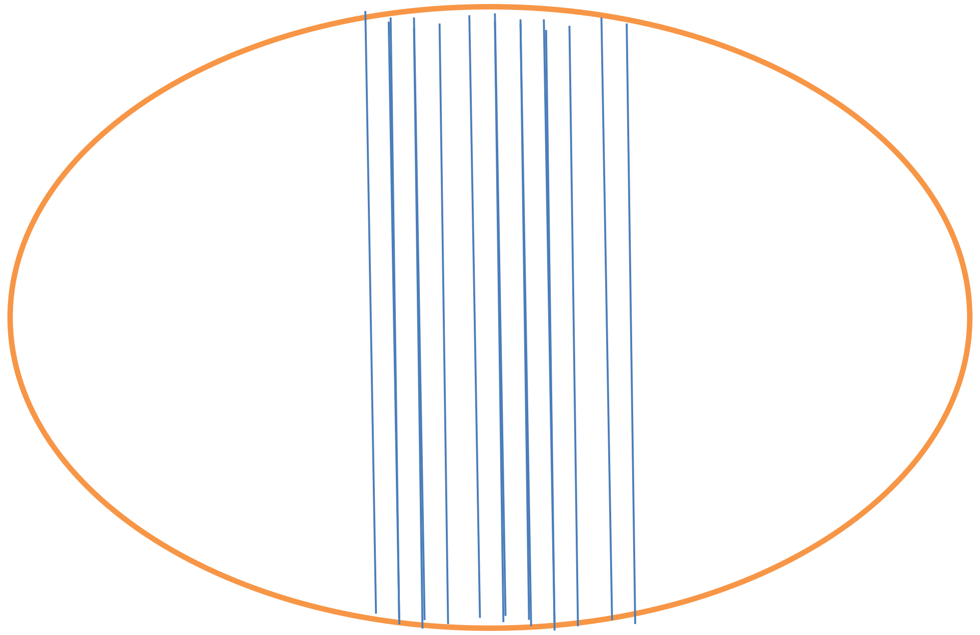
Anders Ahnesjö

Pencil beam



Pencil beam model

Convolution of pencil beams (very small beams) over cross section of beam



Main limitations (approximations) of the model

- Pencil Beam dose distribution is generated in semi-infinite Phantom; patient is nether semi-nfinite: overestimation of scatte dose
- Inhomogeneity correction factors have to be applied - approximation

CPE inhomogeneous aborbent

correction factor for inhomogeneities

Dose distribution is calculated in water equivalent homogeneous absorbent and dose distribution is corrected for inhomogeneites

Definition of correction factor ICF

$$\mathrm{Dose}_{\mathrm{inh}} = \mathrm{ICF} \cdot \mathrm{Dose}_{\mathrm{homogeneous}}$$

Inhomegneity Correction Factor CPE

$$ICF = e^{-\mu \cdot (d_{rad} - d)}$$

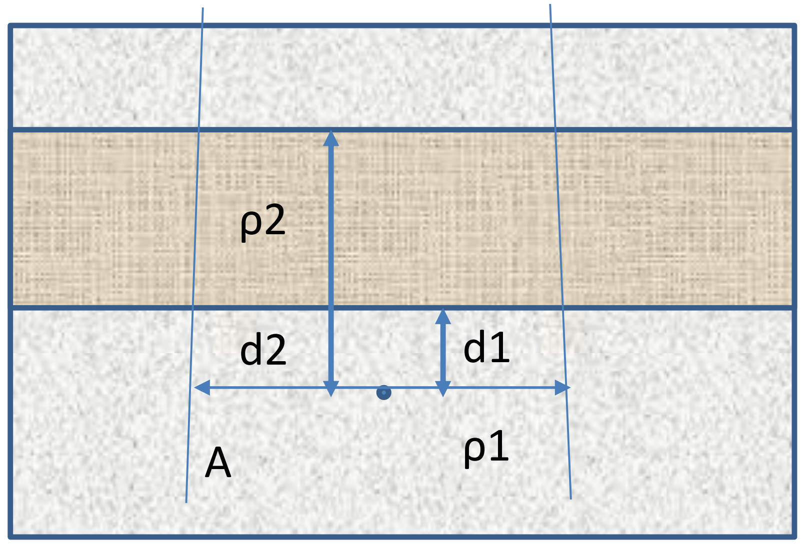
Primary dose changes is accounted for only

$$ICF = \frac{TAR(A, d_{rad})}{TAR(A, d)} = \frac{DD(A, d_{rad}, SSD)}{DD(A, d, SSD)} \cdot \frac{(SSD + d_{rad})^2}{(SSD + d)^2}$$

Primary and scattered dose (???) changes are accounted for

$$ICF = \frac{TAR(A, d_1)^{\rho 1}}{TAR(A, d_2)^{\rho 2}}$$

Batho method
Primary and scattered dose
(?) changes are accounted for



There are other CF, better but complicated example ETAR

Inhomogeneites correction factor

Presence of tissues with composition and density different from water

- lungs,
- bones,
- fat,
- air.

Presence of artificial materials in the body, e.g.

- prosthesis,
- filings (teeth seals).

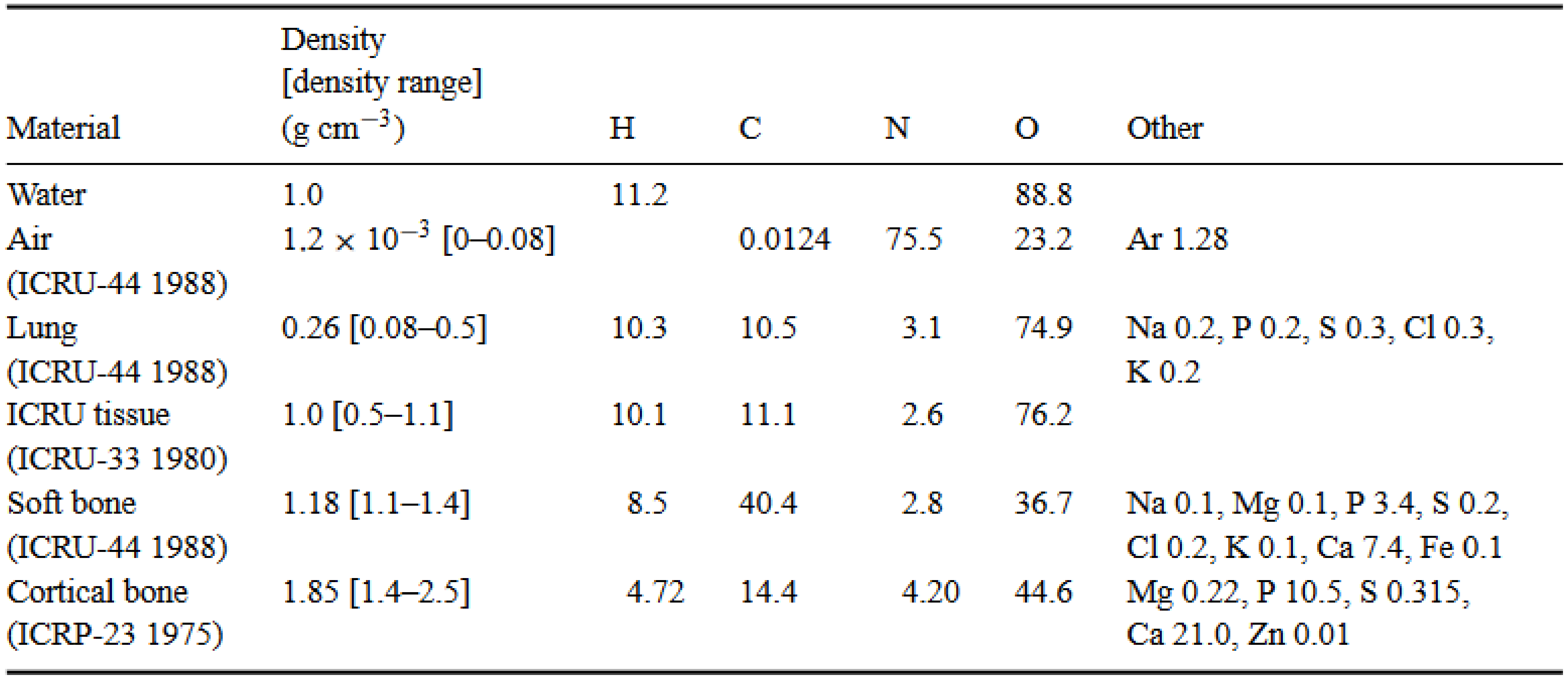
What affects the distribution of the dose in a real situation?

The regions where there is no Transient Charged Equilibrium

- build-up region,
- exit build-down region,
- interfaces regions
  - lung – soft tissue,
  - air – soft tissue,
  - bone – soft tissue,
  - high-Z materials – other tissues.

Range of electrons!

Characteristic of different materials (tissues)



Material	Density [density range] (g cm^{-3})	H	C	N	O	Other
Water	1.0	11.2			88.8	
Air (ICRU-44 1988)	1.2×10^{-3} [0–0.08]		0.0124	75.5	23.2	Ar 1.28
Lung (ICRU-44 1988)	0.26 [0.08–0.5]	10.3	10.5	3.1	74.9	Na 0.2, P 0.2, S 0.3, Cl 0.3, K 0.2
ICRU tissue (ICRU-33 1980)	1.0 [0.5–1.1]	10.1	11.1	2.6	76.2	
Soft bone (ICRU-44 1988)	1.18 [1.1–1.4]	8.5	40.4	2.8	36.7	Na 0.1, Mg 0.1, P 3.4, S 0.2, Cl 0.2, K 0.1, Ca 7.4, Fe 0.1
Cortical bone (ICRP-23 1975)	1.85 [1.4–2.5]	4.72	14.4	4.20	44.6	Mg 0.22, P 10.5, S 0.315, Ca 21.0, Zn 0.01

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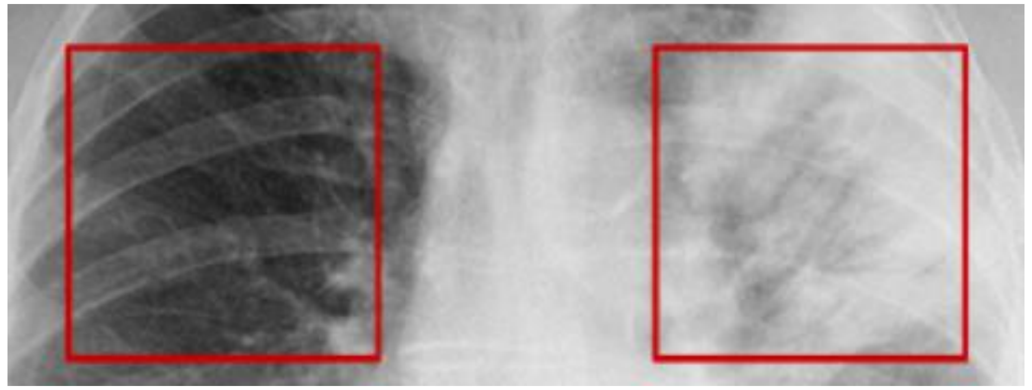
Lung

Like soft tissue composition
inspiration
Upper 0.123 +/- 0.46 g/cm3
Middle 0.121 +/- 0.033
Lower 0.154 +/- 0.057 g/cm3

expiration
Upper 0.215 +/- 0.058 g/cm3
Middle 0.228 +/- 0.066
Lower 0.260 +/- 0.078 g/cm3



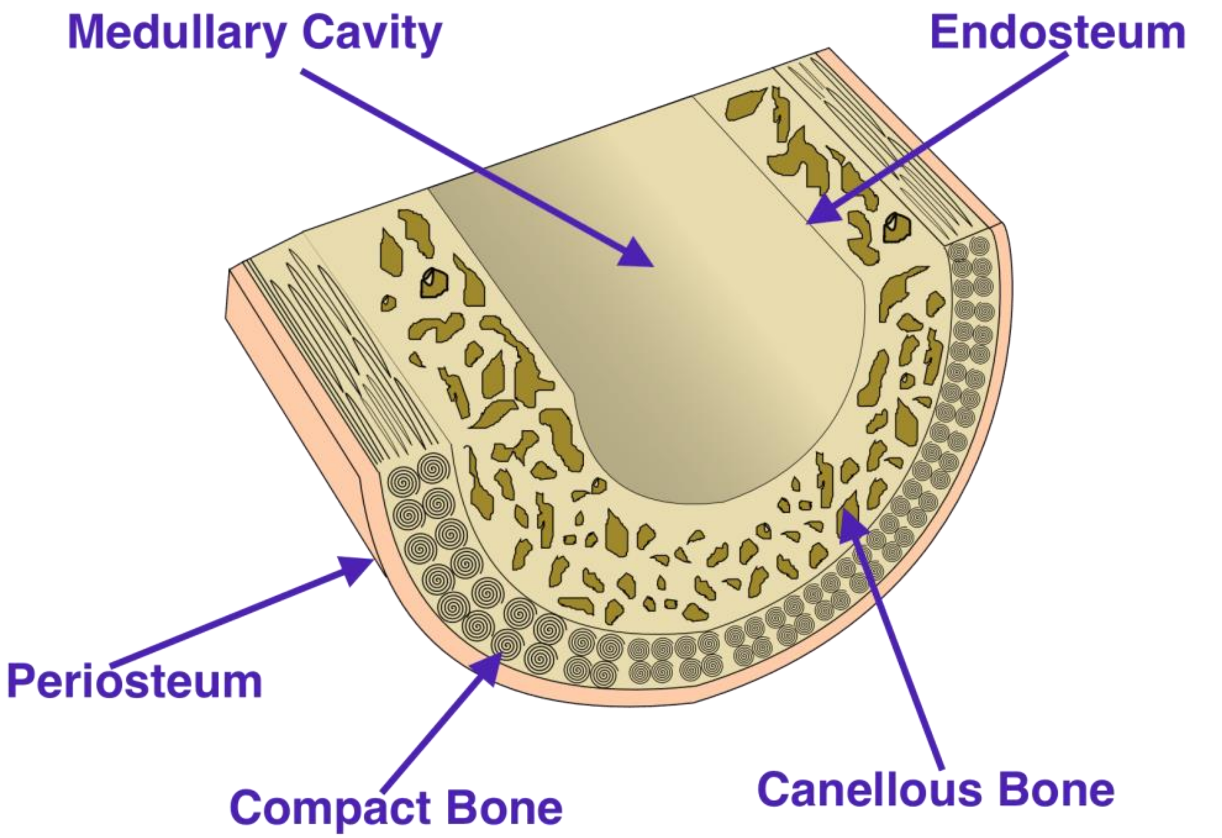
lung abnormalities

Bones

very complicated structure

Compact bone – it is not like soft tissue composition

Calcium
Magnesium
Phosphorus
density 1.6 g/cm^3



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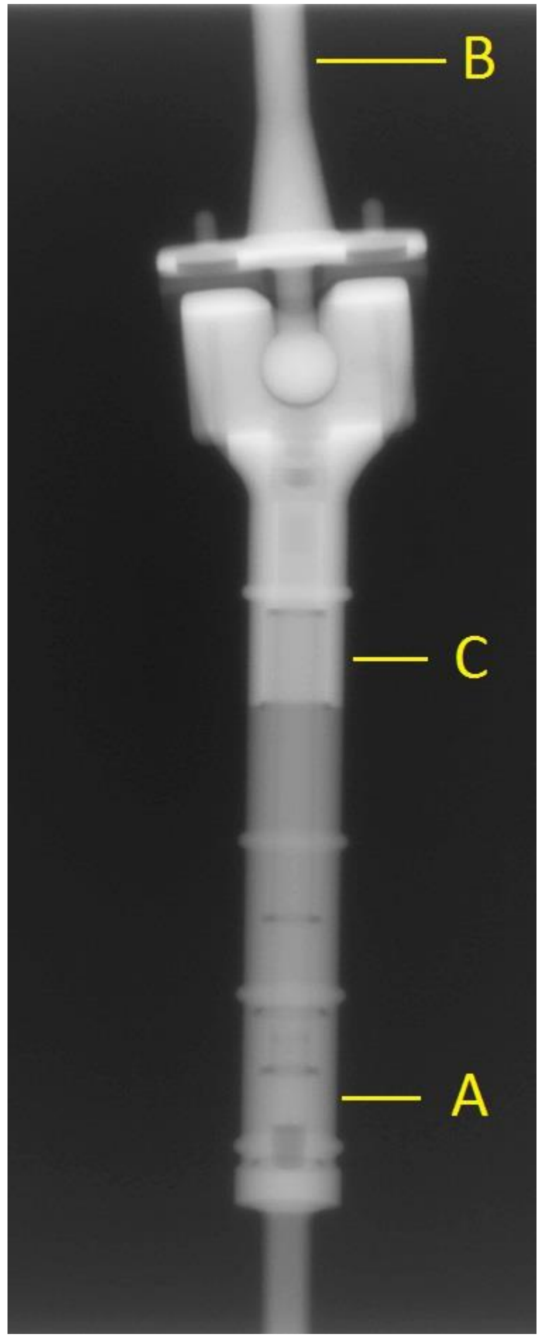
Hip prosthesis



	Co-Cr-Mo alloy	titanium	steel
atomic composition	Co 60% Cr 30% Mo 5%	Ti 90% Al 6% Va 4%	Fe 65% Cr 18% Ni 12 Mo 3
ρ [g/cm^3]	7.9	4.3	8.1
relative electron density	6.8	3.6	6.7

Models used in contemporary TPSs

Convolution models

- ~~pencil beam~~
- colapse cone convolution
- AAA

Monte Carlo models

- electron beams
- for photons at least some elements of Monte Carlo code are used

Acuros – transport of Energy described with Boltzman transport model

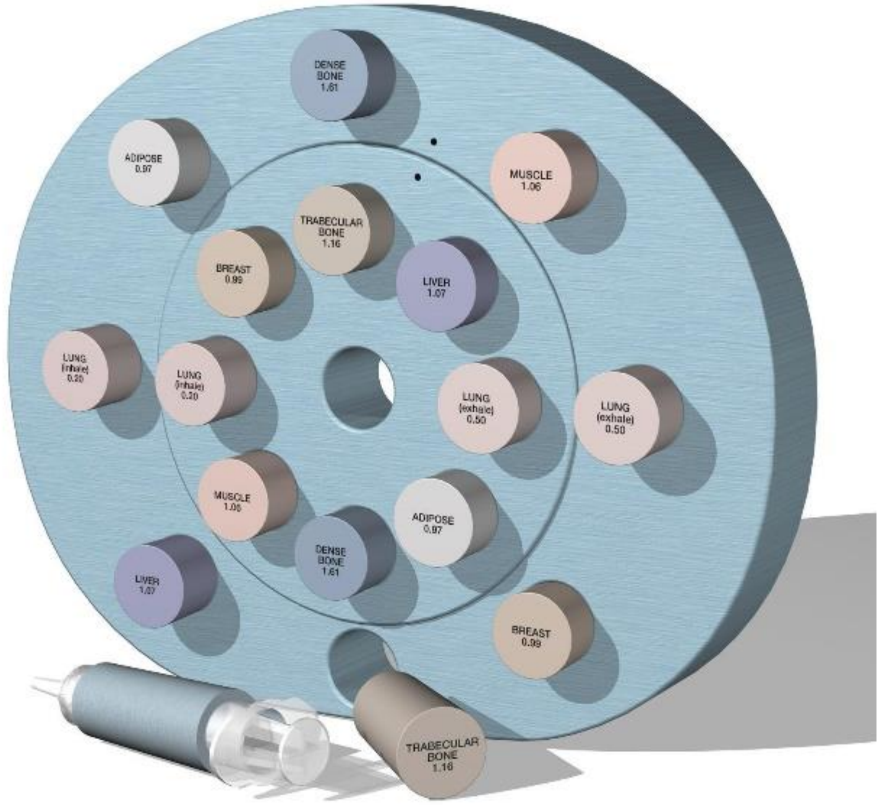
- Varian Eclipse treatment planning system

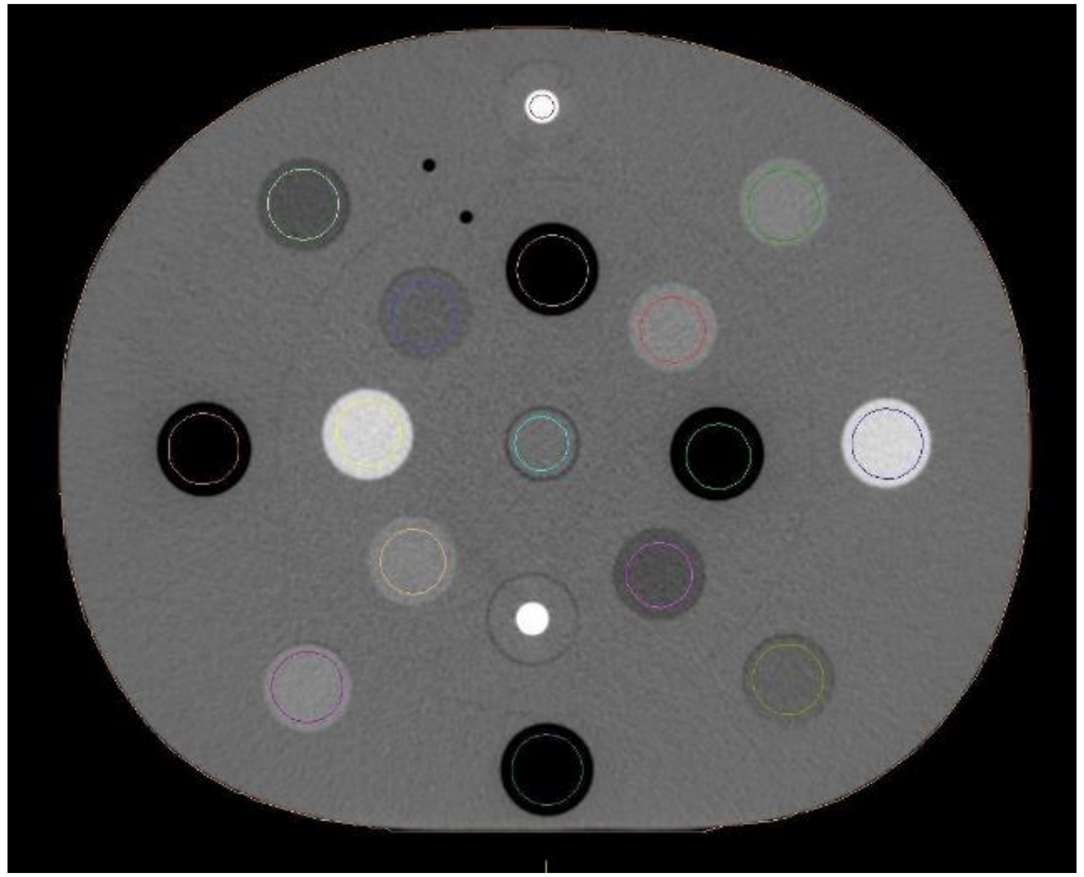
Linear attenuation coefficient is of special importance

$$\Phi = \Phi_{air}^{F} \cdot \frac{F^2}{(F+f)^2} \cdot e^{\sum -\mu_i di}$$

HU – electron density curve measurement HU-ED

e.g. CIRS Phantom
- special H-Z inserts
  - aluminium, brass, steel





If you use different CT protocols
Measure HU-ED curve
for all of them

Collapsed Cone Convolution

acceleration of computation- approximation

30 x 30 x 30 cm^3 water phantom
0.3 cm calculation grid

100 x 100 x 100 calculation points 10^6

Real full convolution: each point with each point
10^6 x 10^6 = 10^{12} too long!

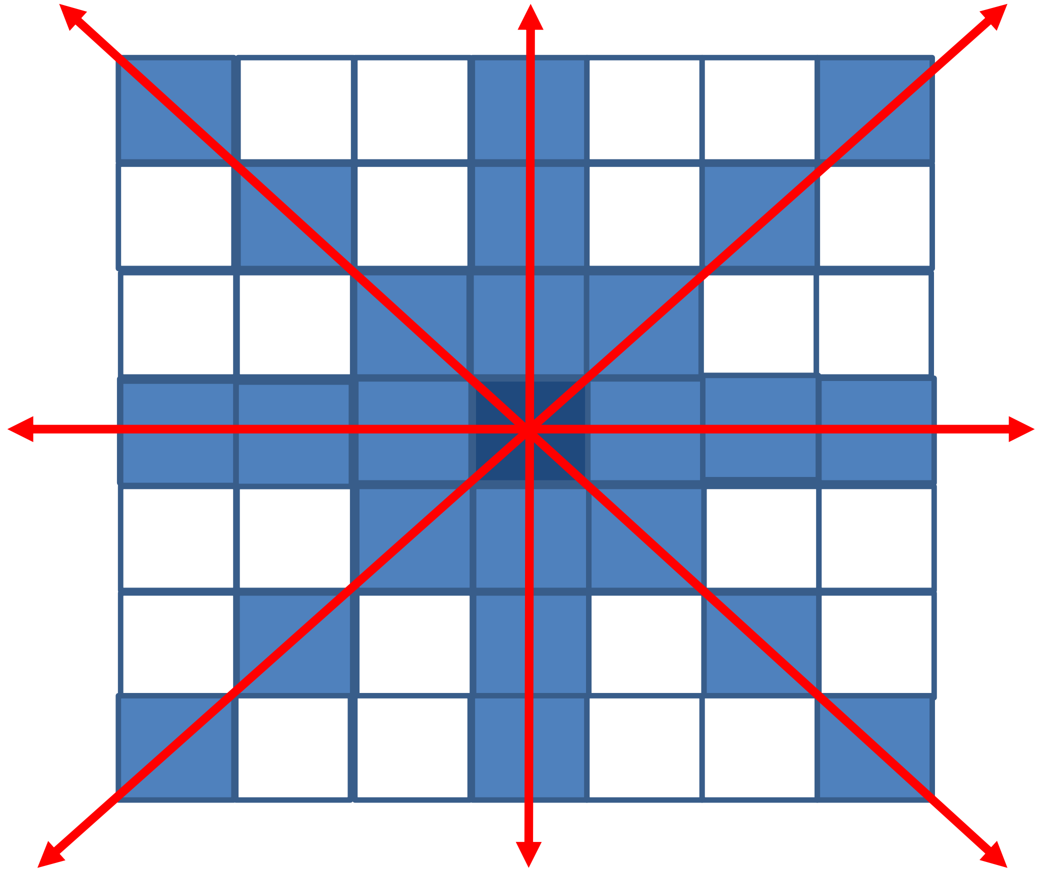
Dose deposition decreses very much with distance!

Mackie’s opinion – 100 cones is enough!

Mobius3D – 144 collapsed cones

Pinnacle – 80 collapsed cones

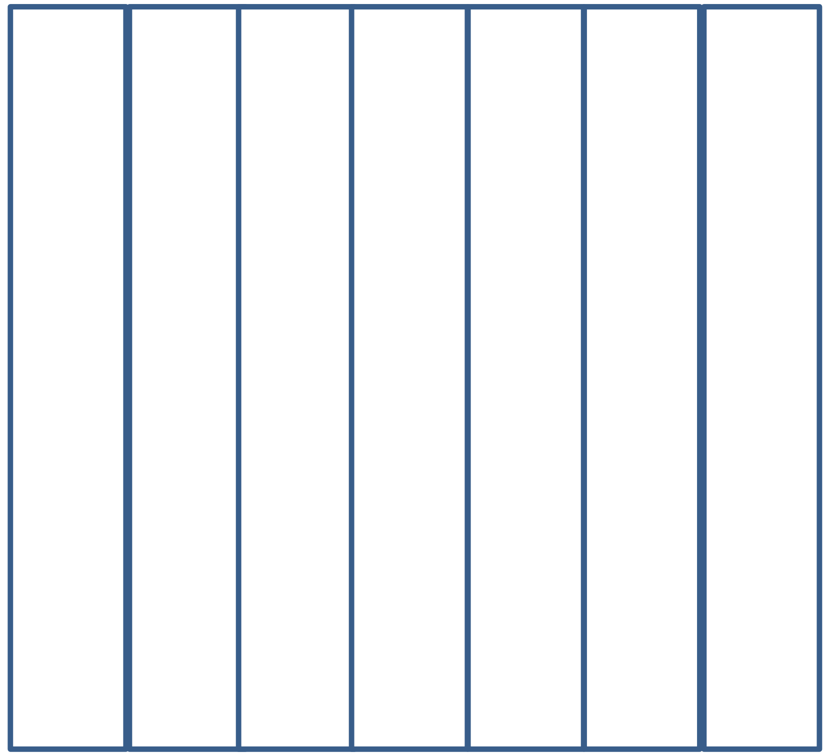
$$D(\bar{r}, h\nu) = \int \frac{dT_{h\nu}(\overline{r'})}{dh\nu} A_{h\nu}(\bar{r} - \bar{r}'; h\nu) d^3\overline{r'} dh\nu$$



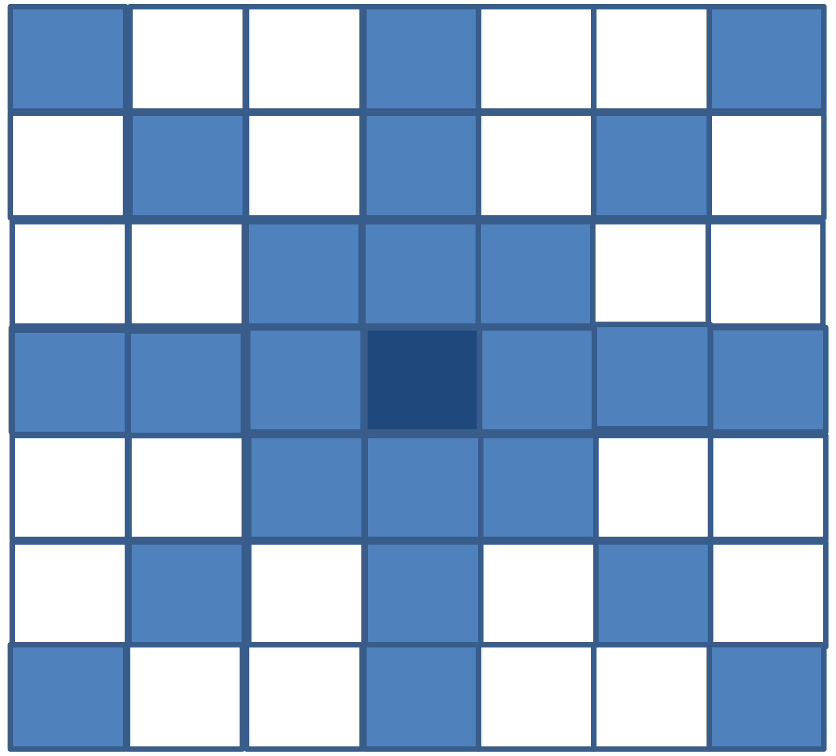
8 cones
Calculations for blue elements only!

Difference between Pencil Beam and Colapse Cone Convolution

Pencil beam



Colapse Cone Convolution



Algorithms

Collapse cone convolution

Eclipse

- Anisotropic Analytical Algorithm

RayStation

Acuros Boltzman’s transport model

Monte Carlo
partly used in all treatment planning systems

Elekta
In Monaco Monte Carlo is used

Penumbra

Penumbra may be defined in terms of

- size of the source and collimator geometry (geometrical penumbra)
  - distance from the source and collimator (small differences for X and Y direction)
- range of electrons (physical penumbra)
  - energy and spectrum of beam

In Eclipse the efective spot size parameters are define by the user

- comparison of calculations with profiles measurements (esspecially important for small beams)

Penumbra

effective spot size

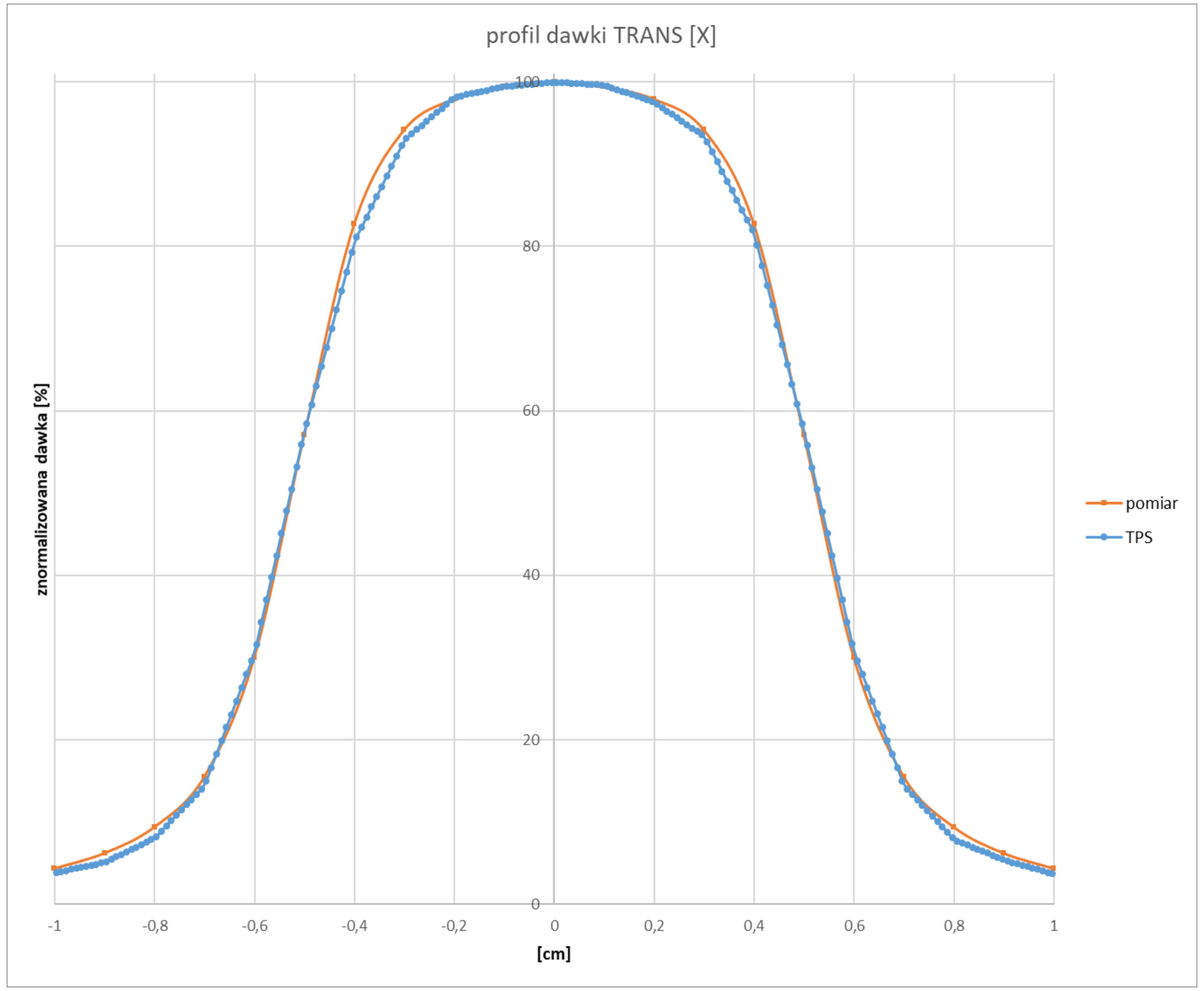
Effective target spot size in X-direction and Y-direction have a significant effect on the calculated absolute dose level for:

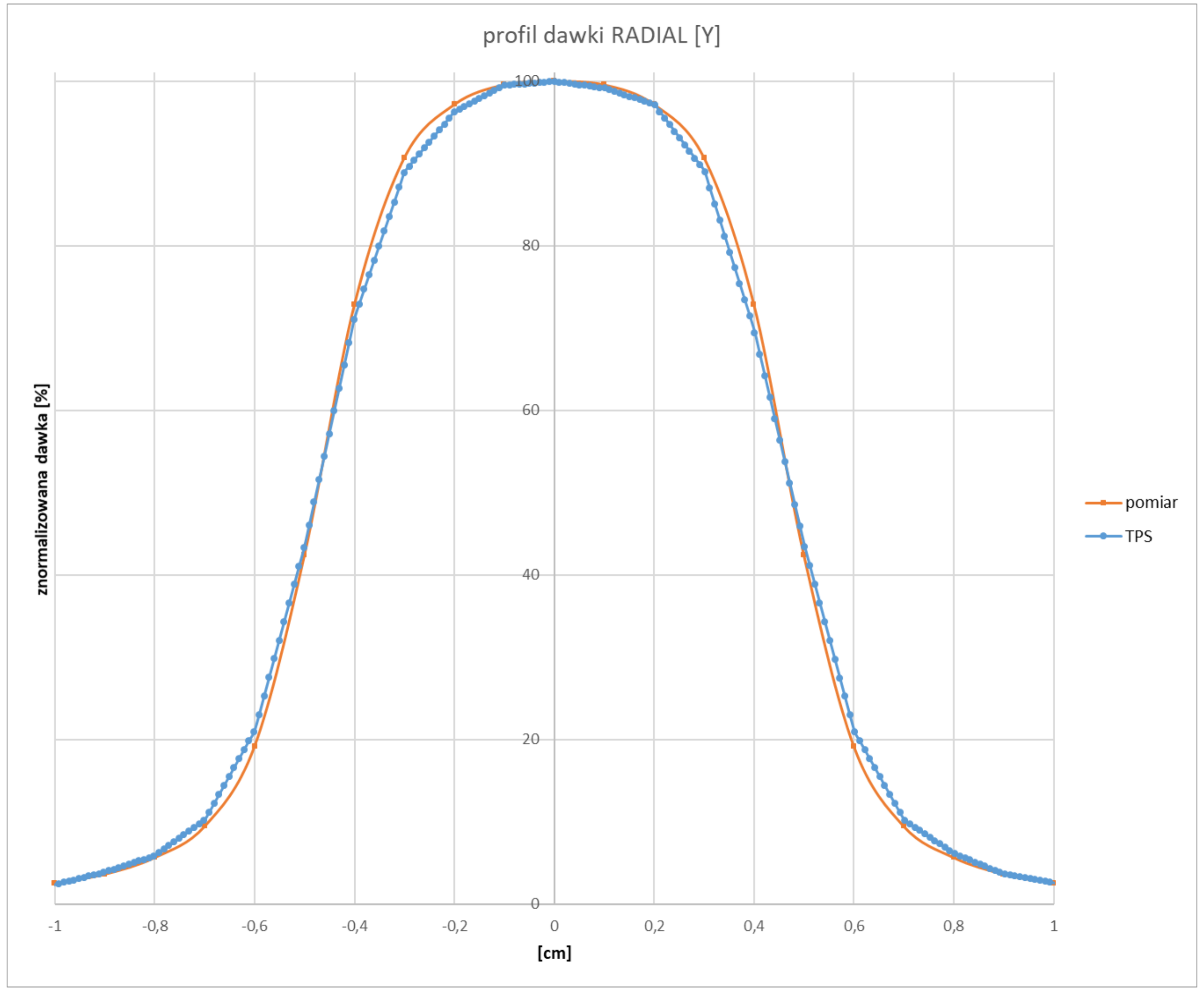
- very small field sizes (<= 1x1 cm^2),
- the shape of the calculated penumbra for all field sizes.

Adjustement procedure based on comparison of measured and calculated profiles.

X direction radial direction

1 x 1 cm^2, source size 1 x 1 mm

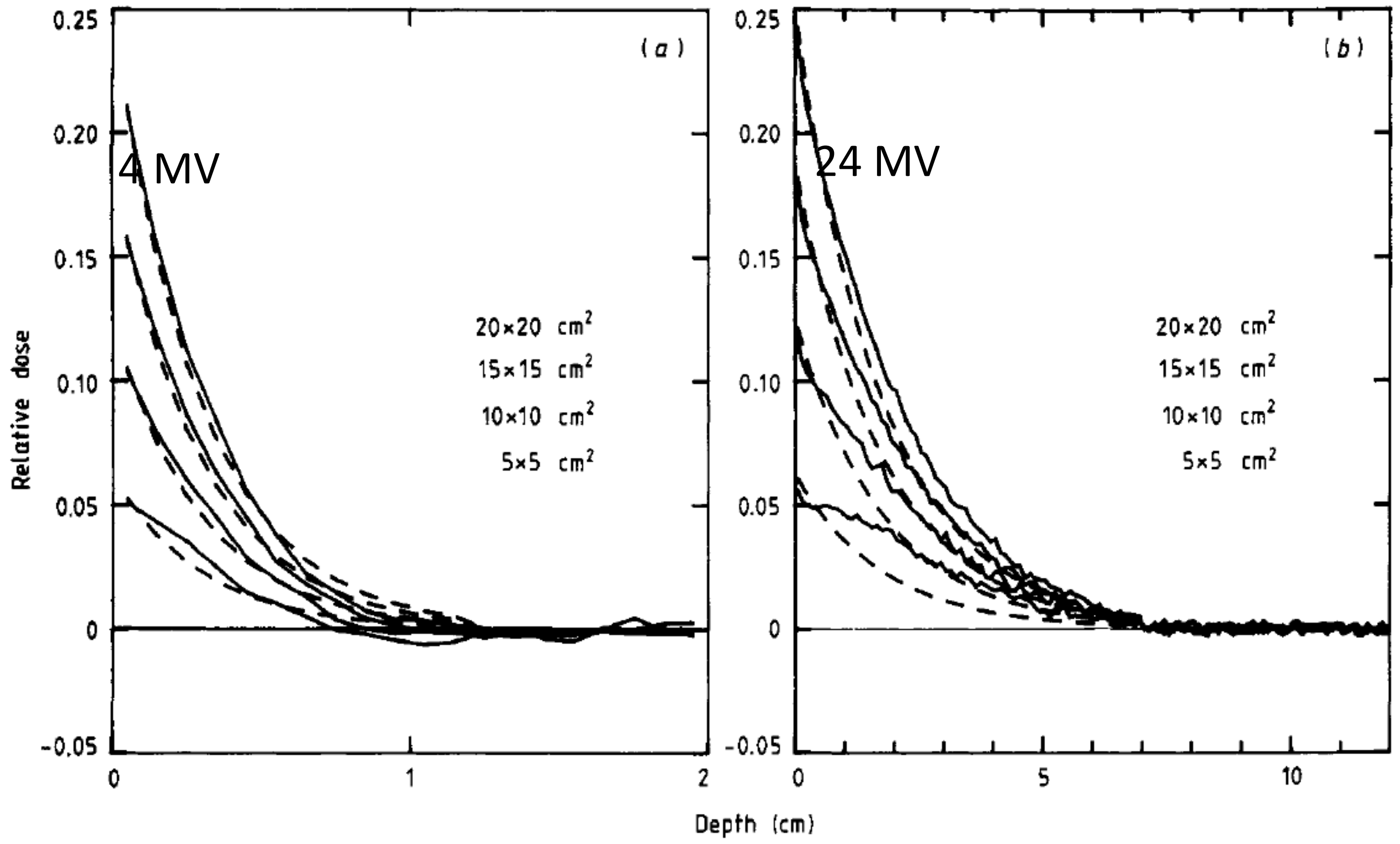




Courtesy of Medical Physics Department, Warsaw

Buil-up dose distribution

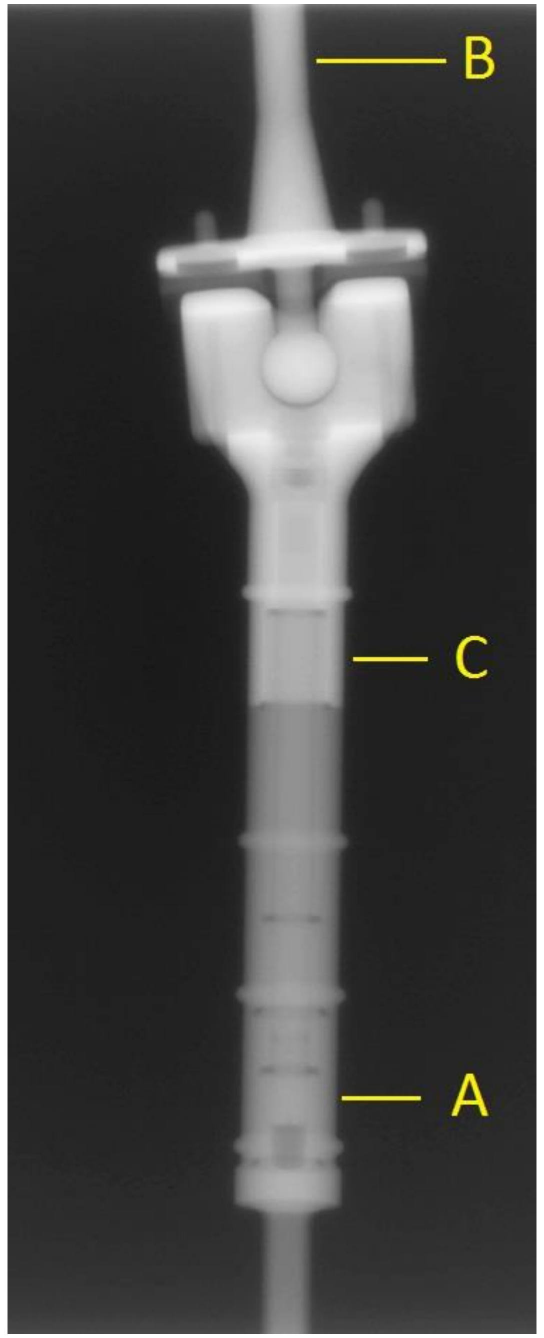
Charged particle contamination was separately accounted for



Fractions of the maximum total dose

Ahnesjo, Andreo, 1989, PMB 34

Hip prosthesis



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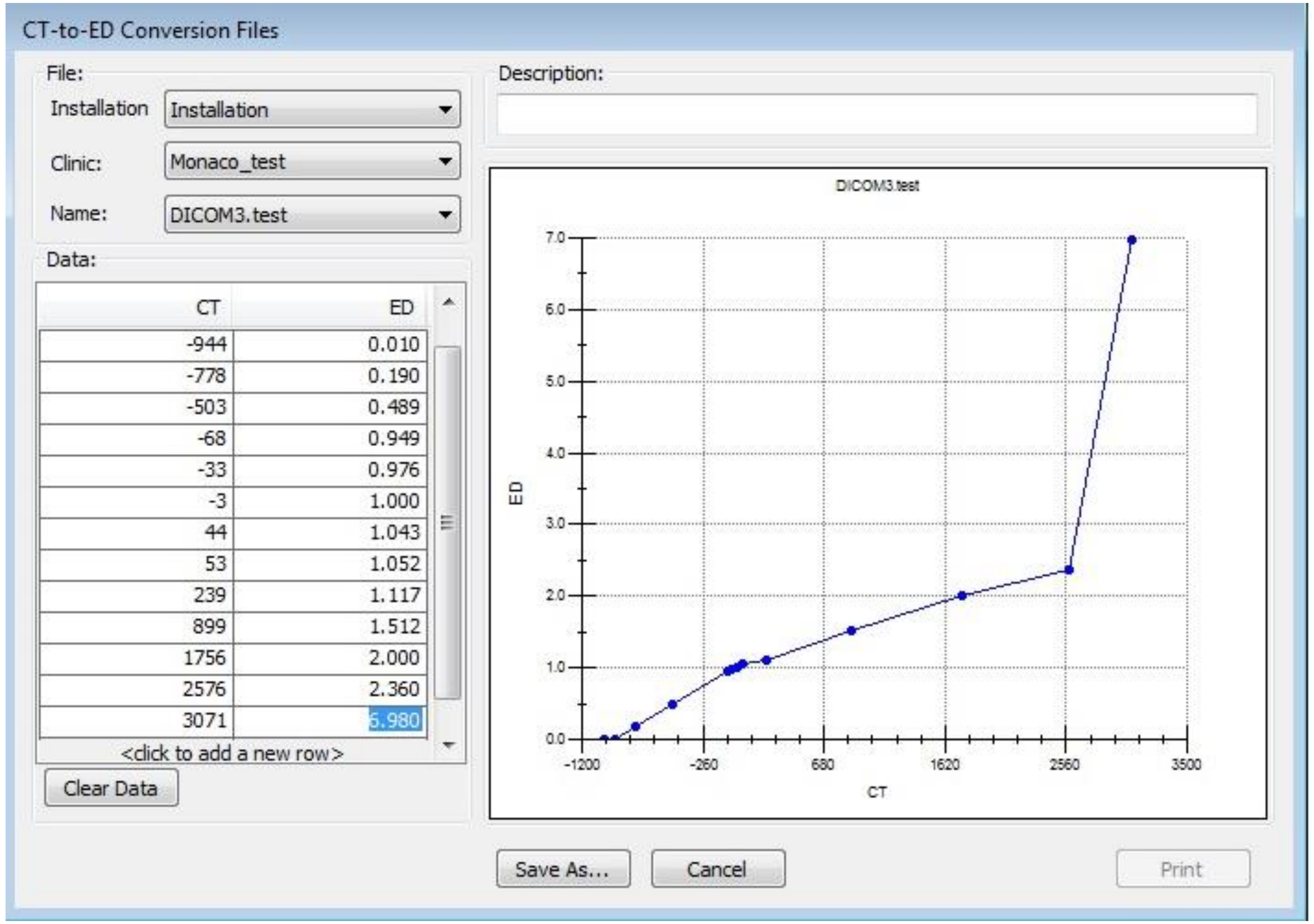
atomic composition	Co 60% Cr 30% Mo 5%	Ti 90% Al 6% Va 4%	Fe 65% Cr 18% Ni 12 Mo 3
ρ [g/cm^3]	7.9	4.3	8.1
relative electron density	6.8	3.6	6.7

HU – electron density conversion curve



Installation: Installation

Clinic: Monaco_test

Name: DICOM3.test

CT	ED
-944	0.010
-778	0.190
-503	0.489
-68	0.949
-33	0.976
-3	1.000
44	1.043
53	1.052
239	1.117
899	1.512
1756	2.000
2576	2.360
3071	6.980

What we should remember of?

Standard mode

- 12 bits up to 2^{12}; 4096 HU: -1204 - +3071 (aluminium)

Extended mode

- 16 bits up to 2^{16}; 65536 HU (any material)

High Z materials and dose distribution

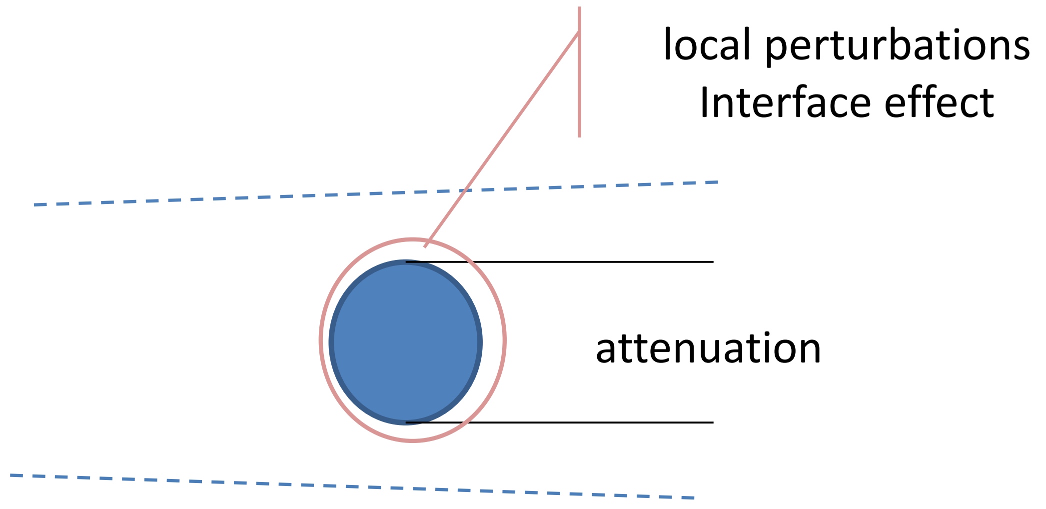
hip prosthesis

Attenuation

energy photon fluence is smaller due to
attenuation of photons dose is smaller

Local perturbations – interface effects

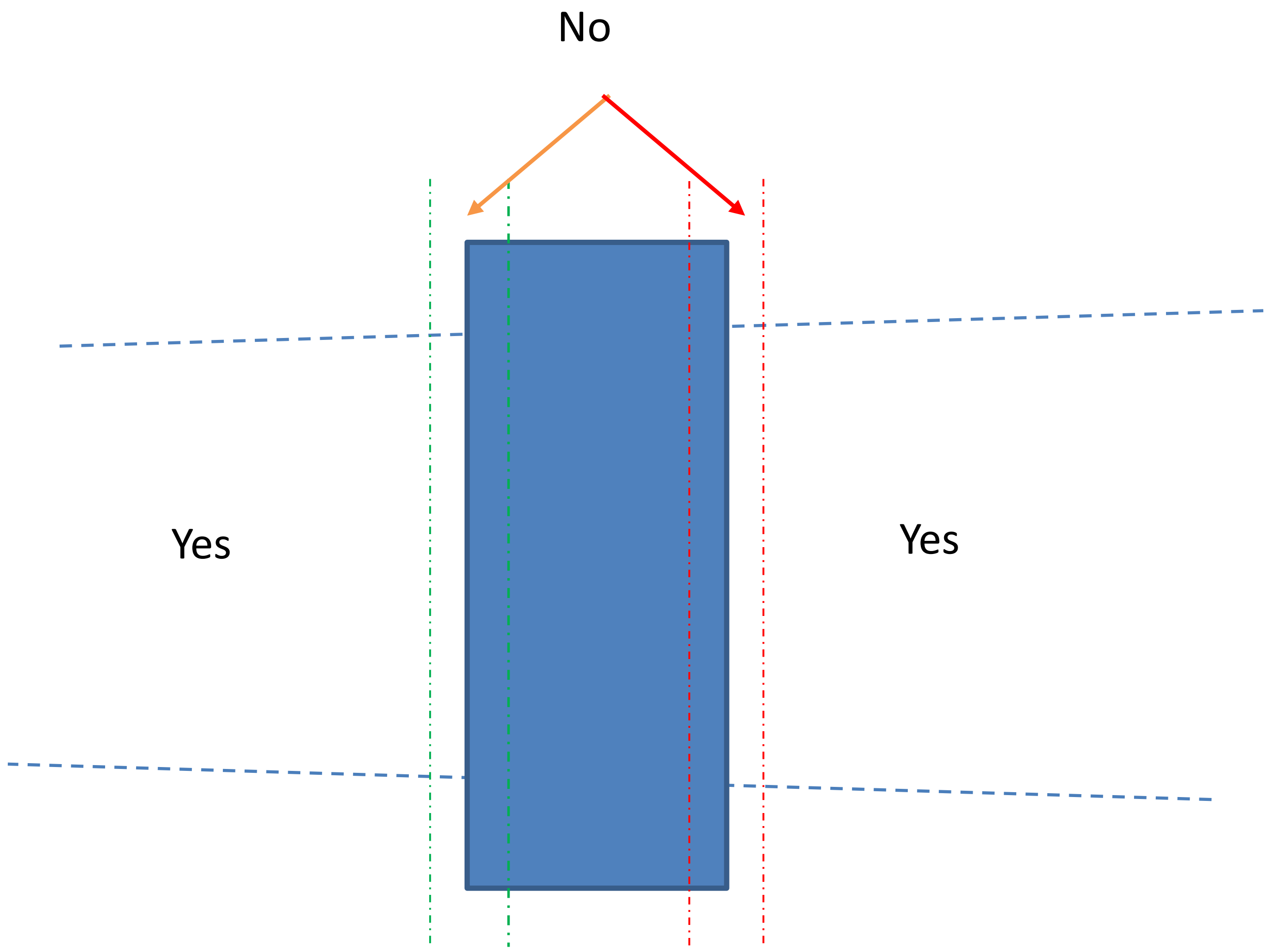
energy electron fluences is changed by local perturbations



Slab geometry

easier to understand

charged particle equilibrium



Back scatter

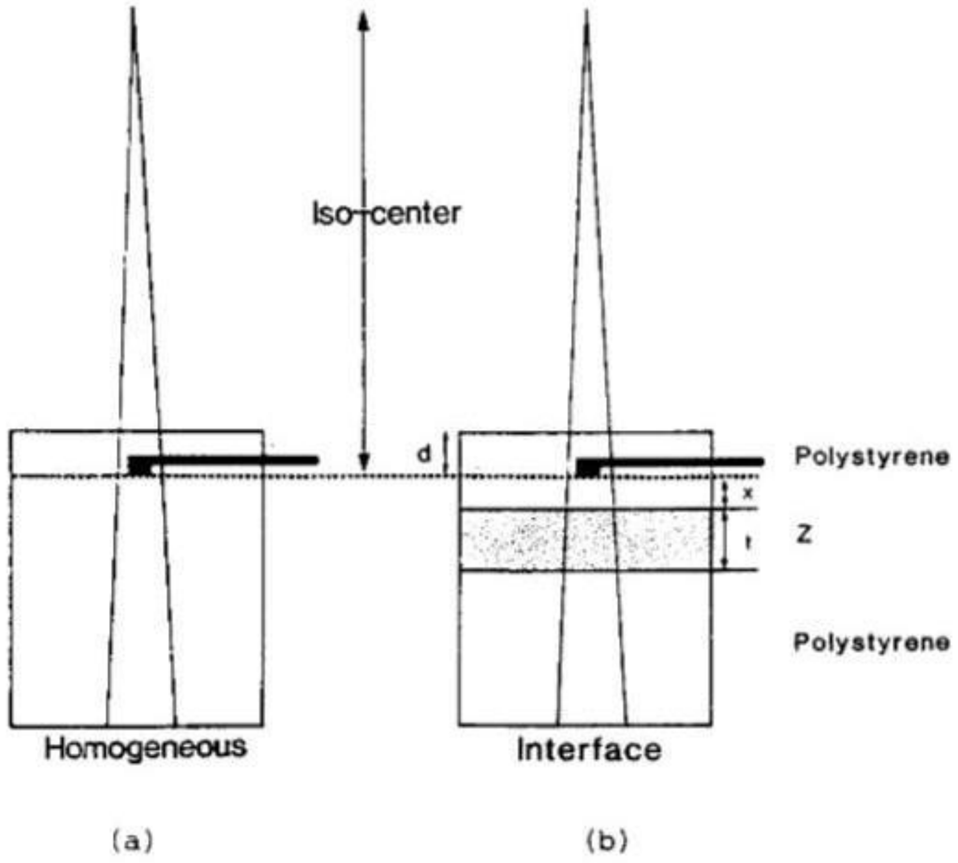


FIG. 2. Experimental setup for the measurement of the backscatter dose factor (BSDF). The ratio of readings in two setups (interface and homogeneous) give the BSDF defined in Eq. (1).

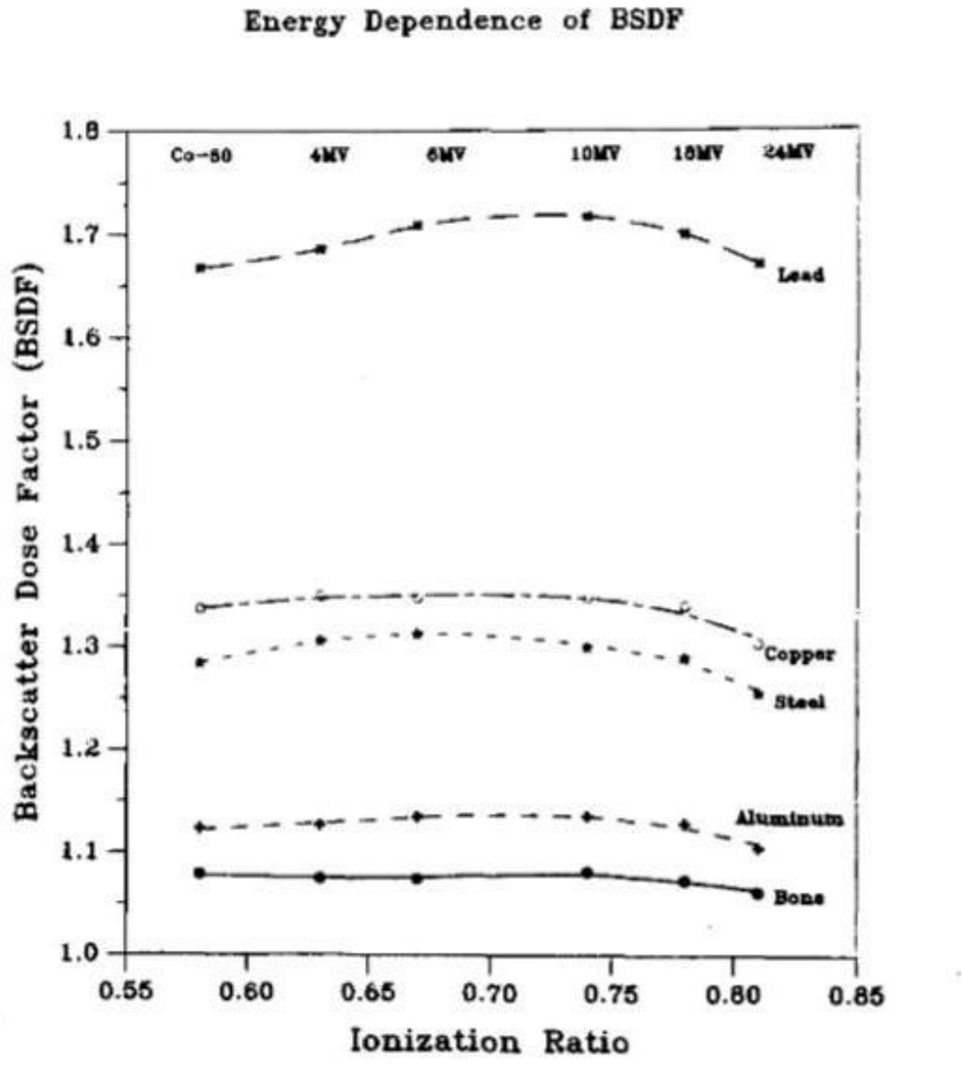
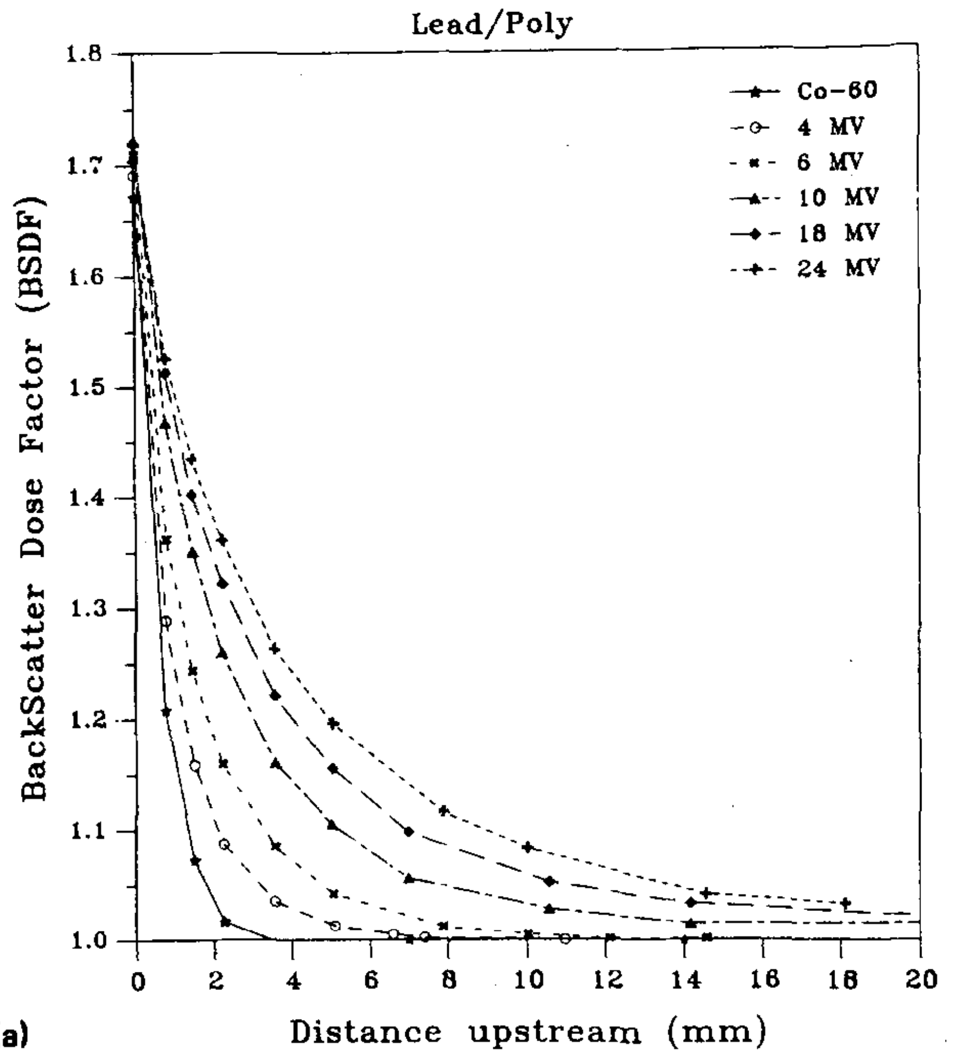


FIG. 5. Backscatter dose factor (BSDF) vs energy of the photon beams plotted as the ionization ratio defined in AAPM Protocol TG-21, for various media.

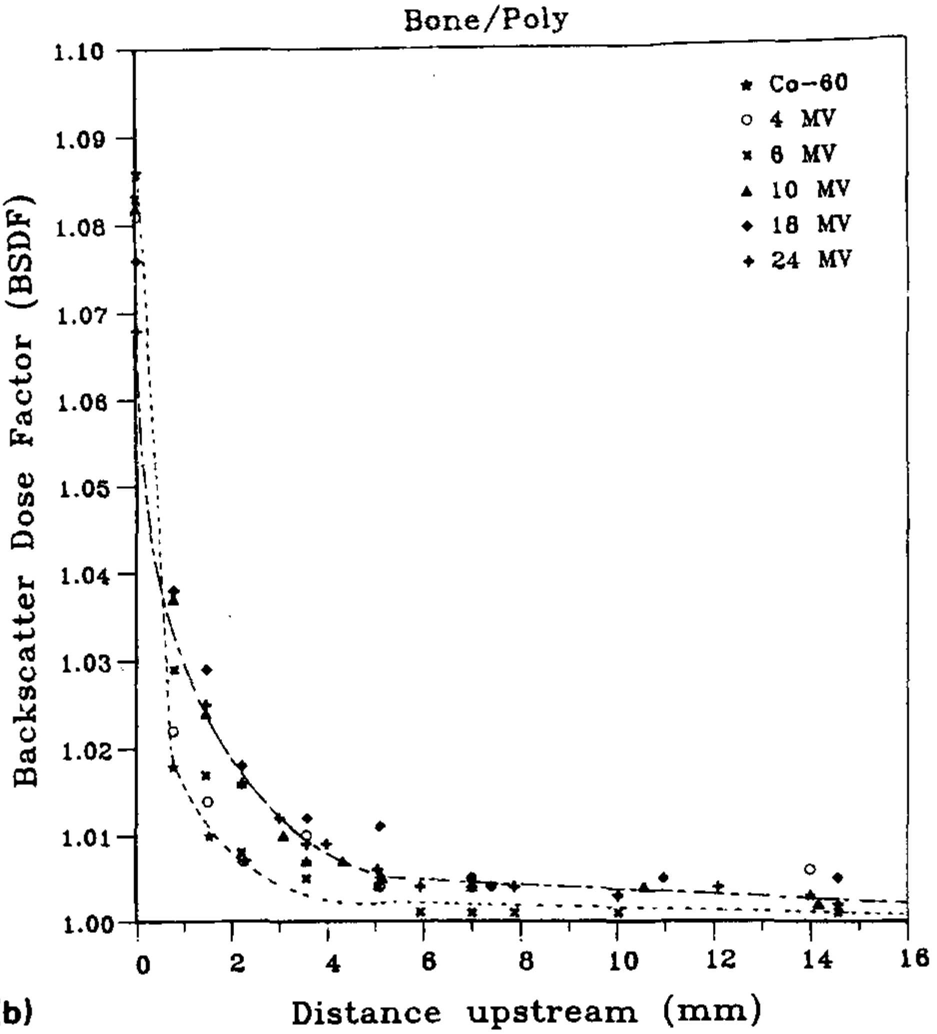
Med. Phys. Das 1989, 16 (3)

Distance upstream

dependece on distance



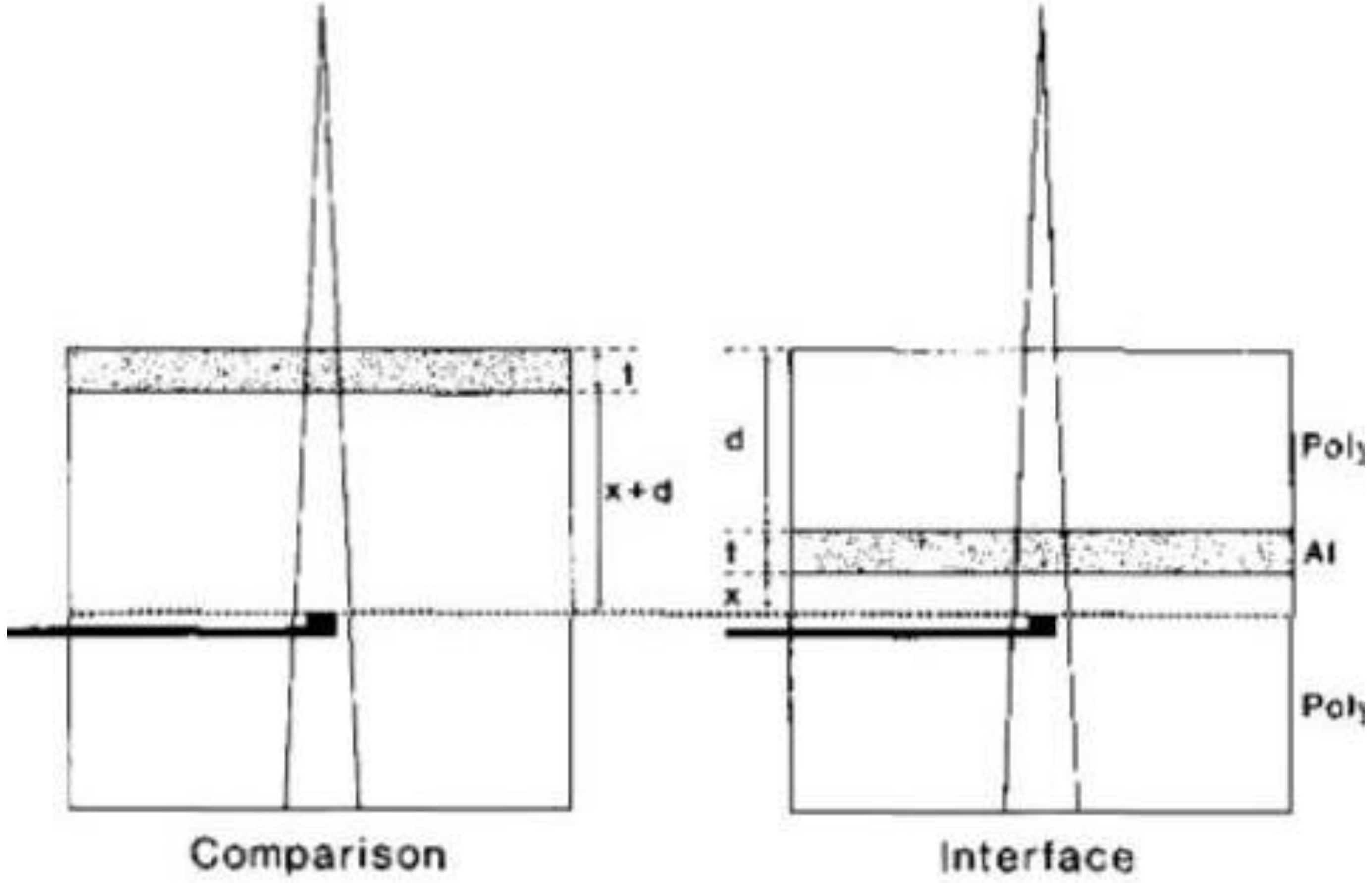
(a)

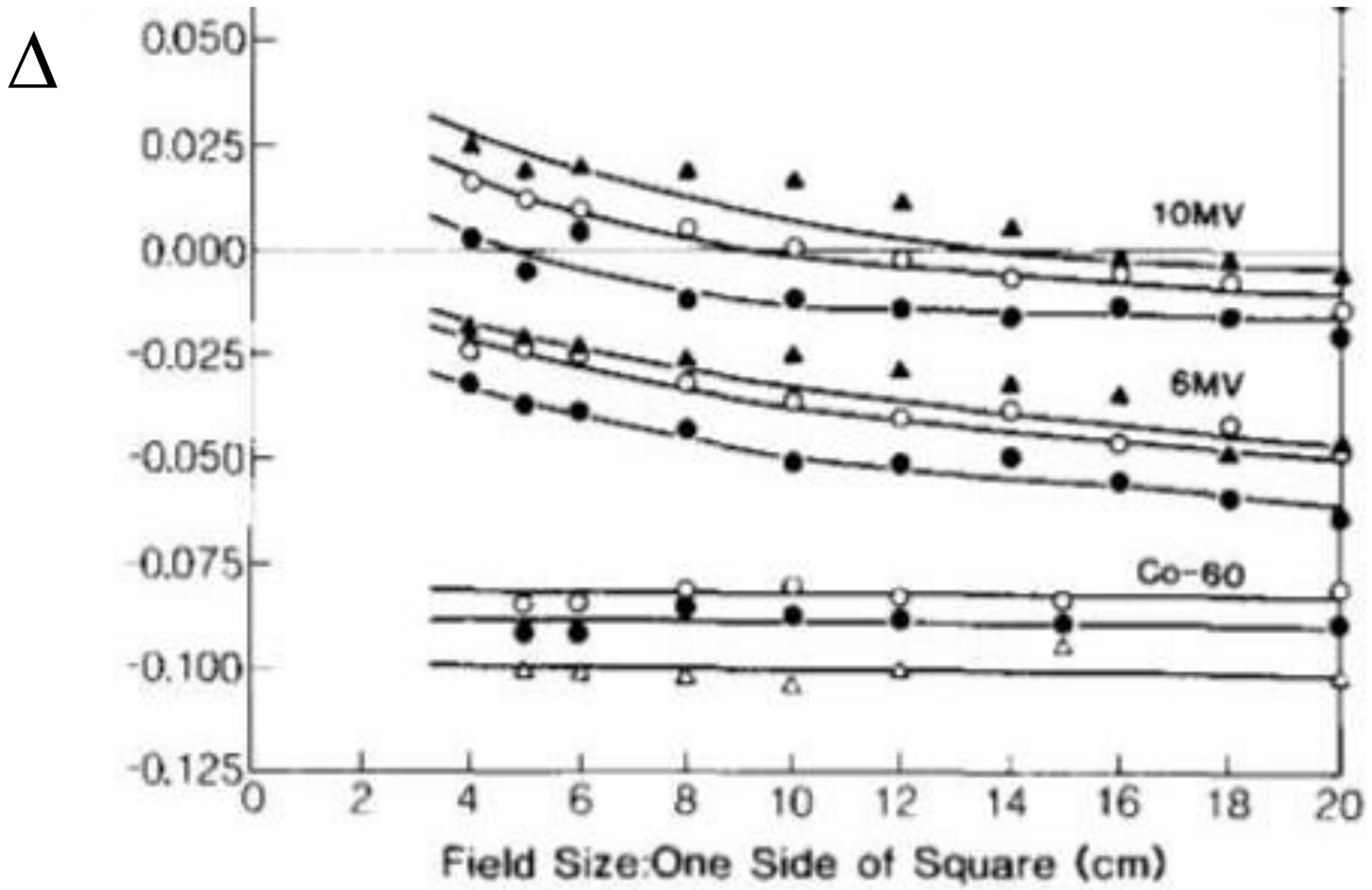


(b)

Downstream dose – corrected for attenuation multiplied by $e^{\mu d}$

$$\Delta = \frac{dose\_at\_\mathrm{int}erface - comparison\_dose}{comparison\_dose}$$

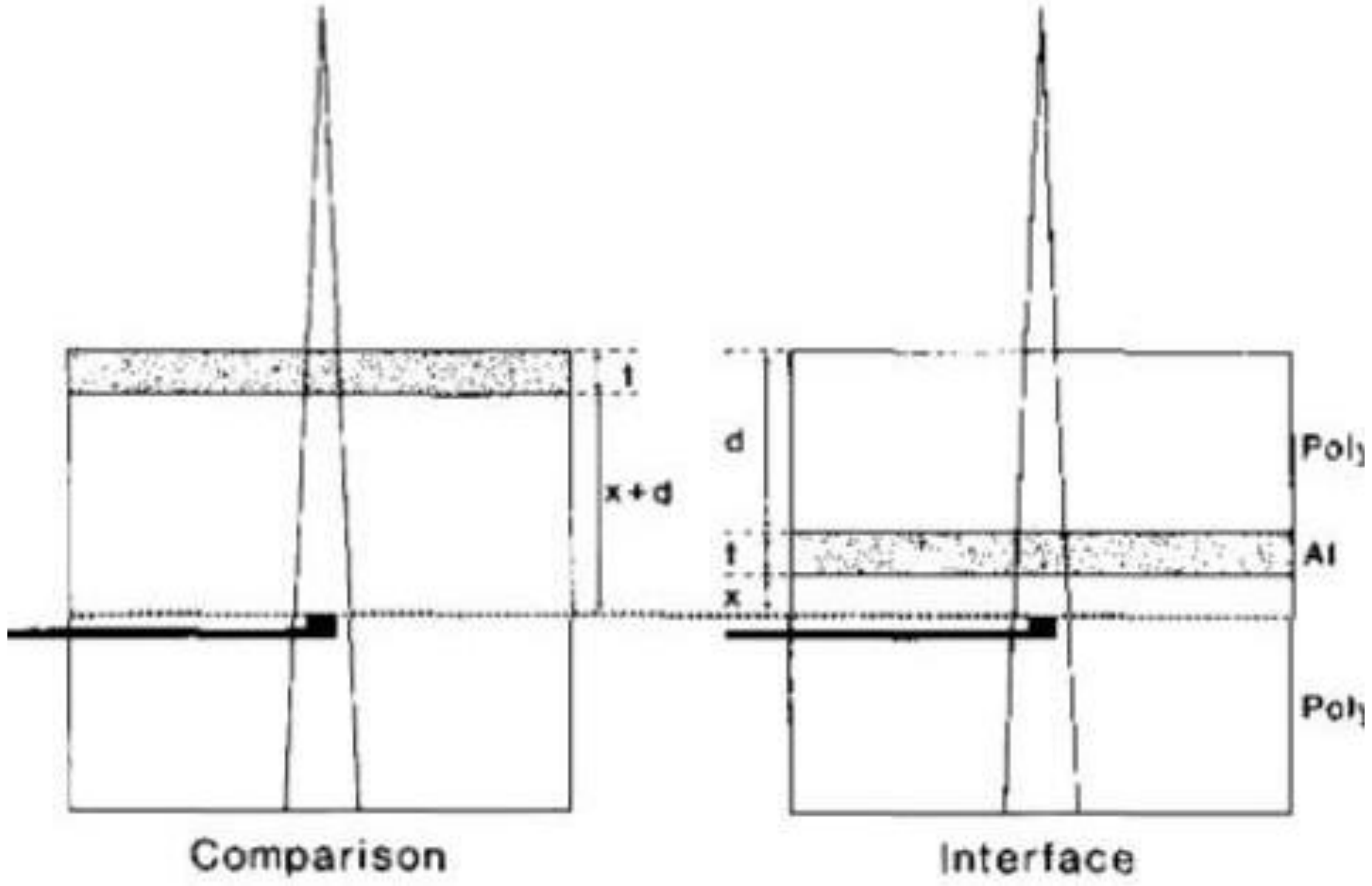


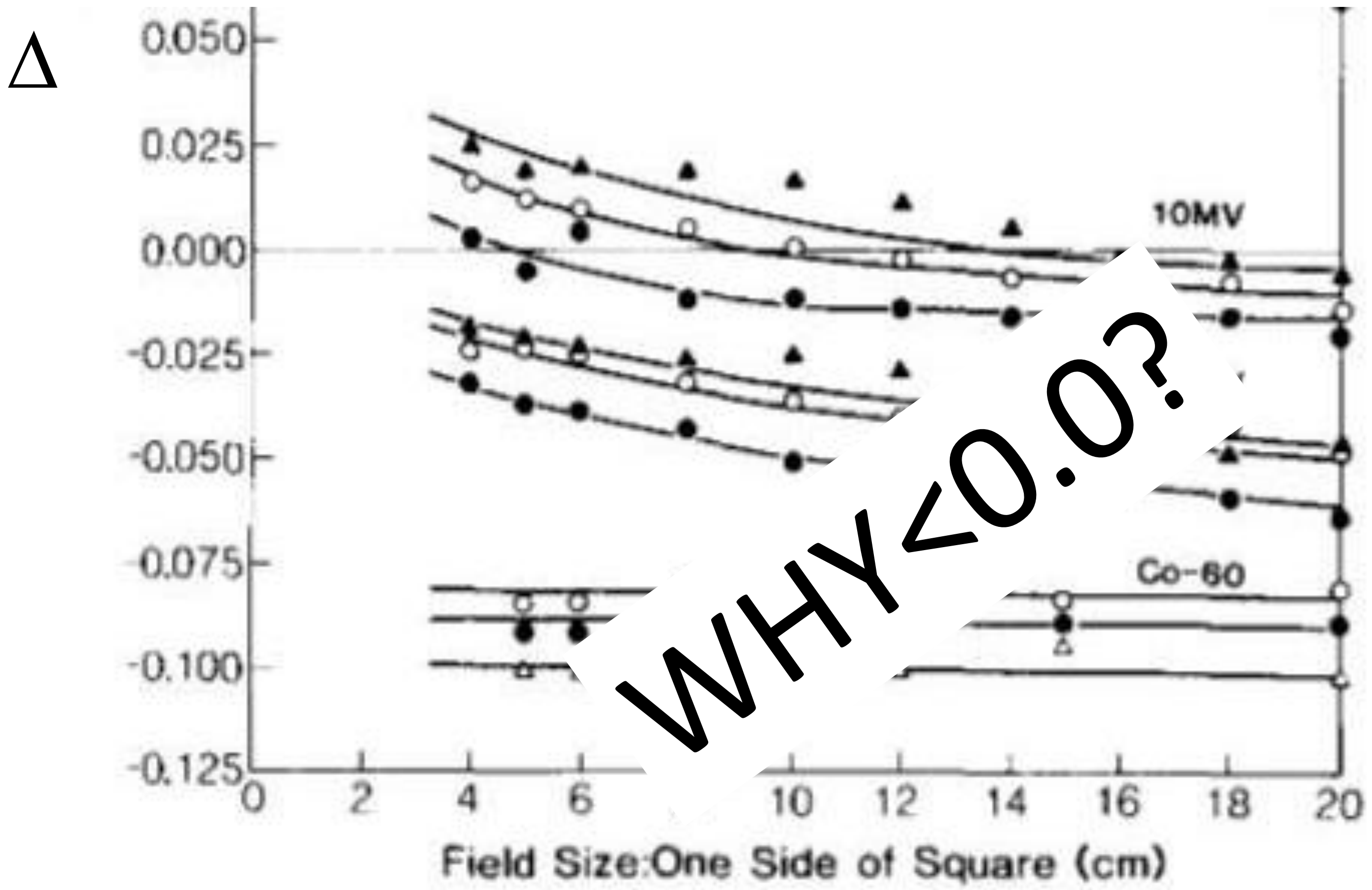


t = 1.2 cm for Co60, 4 cm for other energies

Downstream dose – corrected for attenuation

$$\Delta = \frac{dose\_at\_interface - comparison\_dose}{comparison\_dose}$$



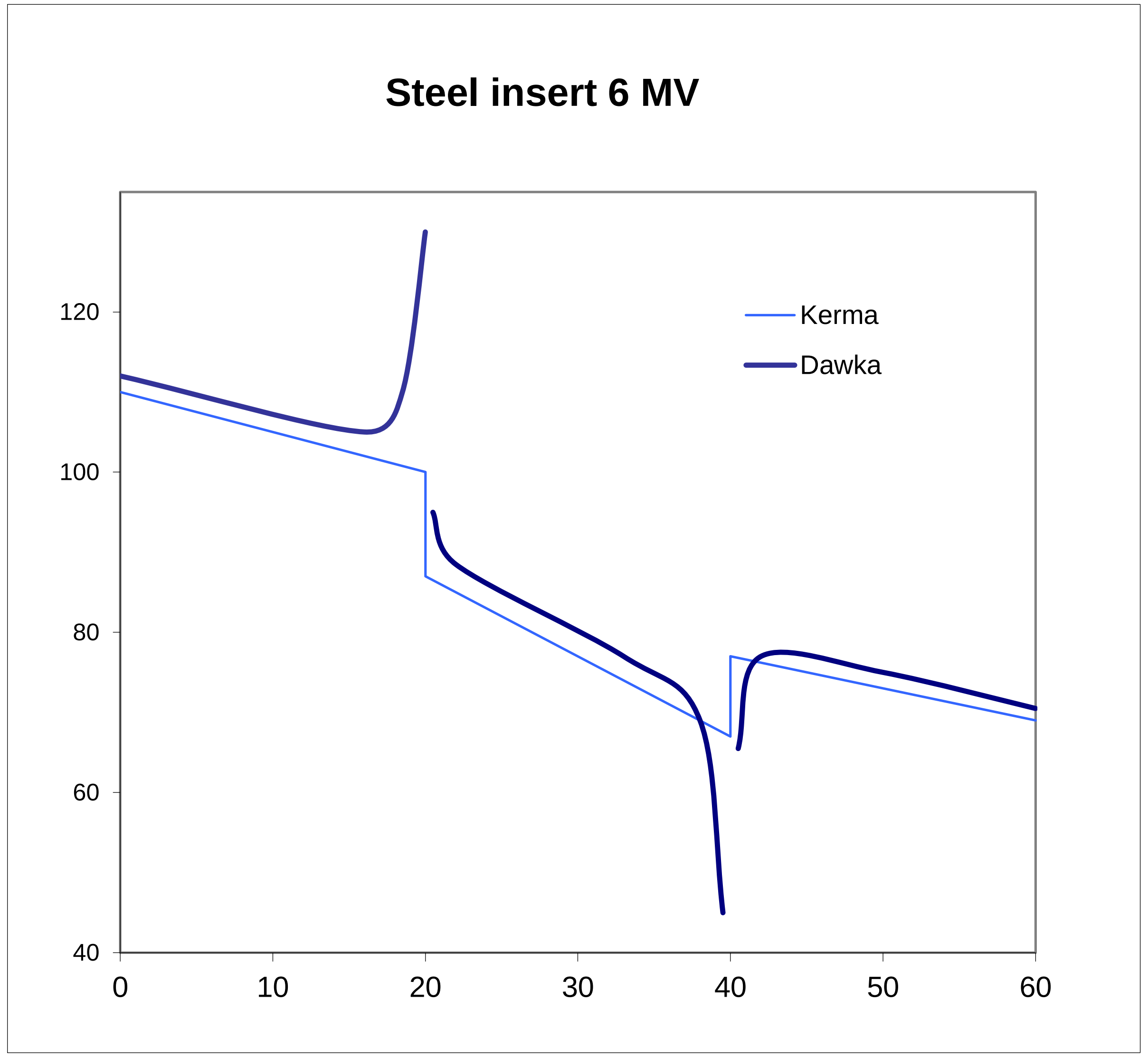


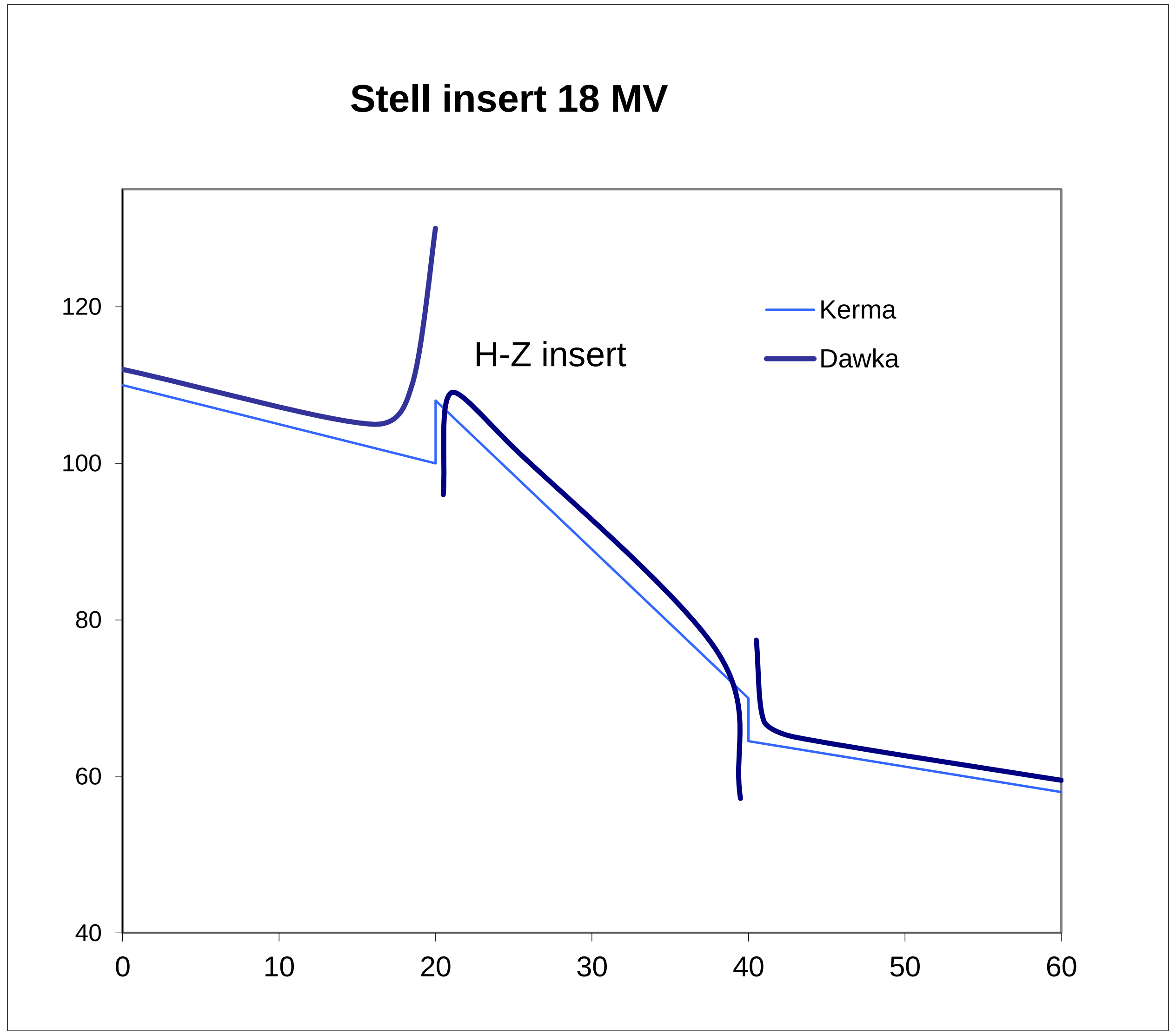
WHY<0.0?

t = 1.2 cm for Co60, 4 cm for other energies

WHY < 0.0 ? (remember: corrected for attenuation)

because dose is deposited by electrons generated in another material
spectrum of electrons is different and mass stoping power ratio is different





Calculation algorithm

In general

- superposition-convolution algorithms give good results in CPE region,
- Monte-Carlo – the only one may calculate very well (RaySearch, Monaco)

  the dose in regin where there is no CPE
- Acuros gives quite good results

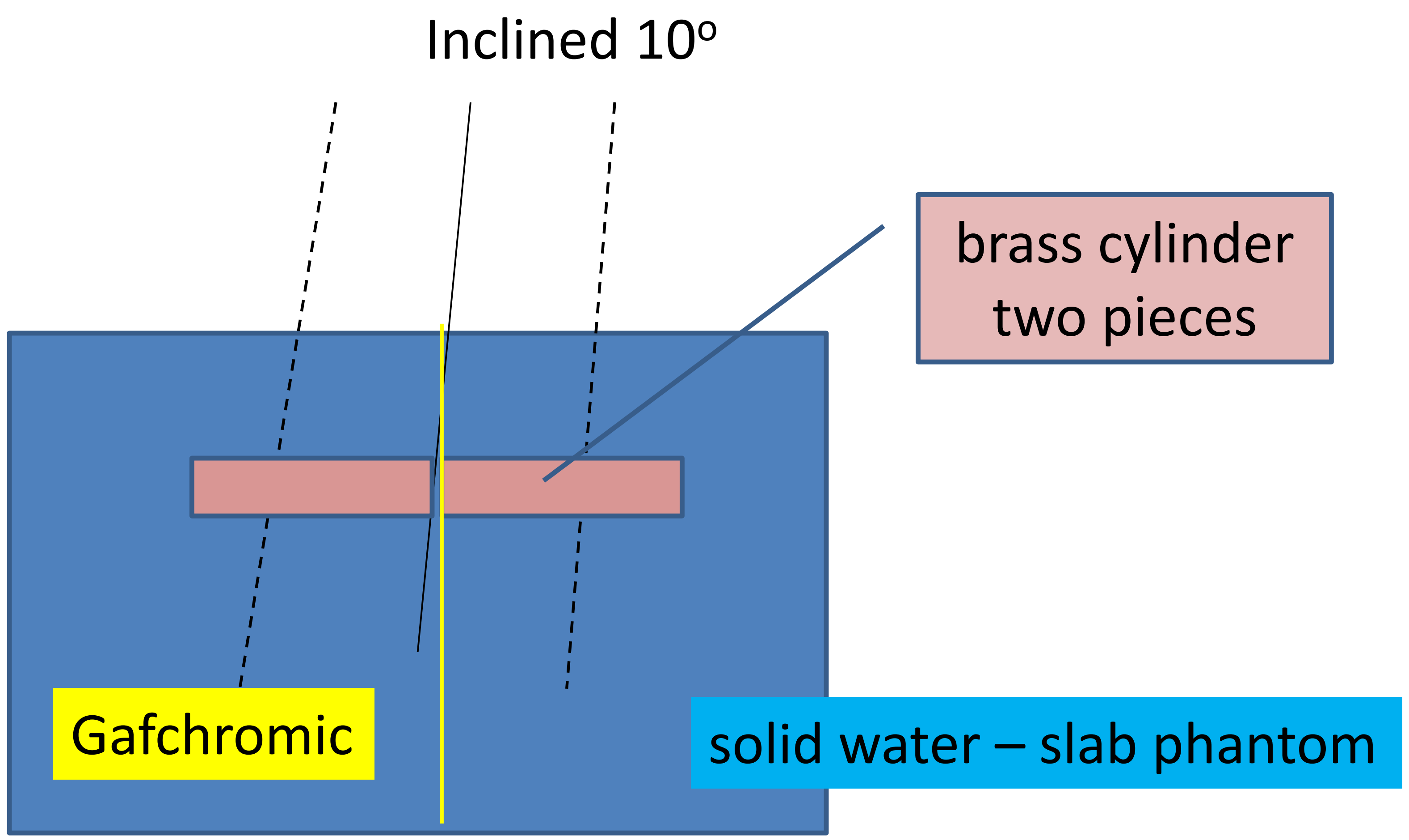
Dose to water dose to medium (tissue)

Bragg-Gray theory – electron fluence is the same

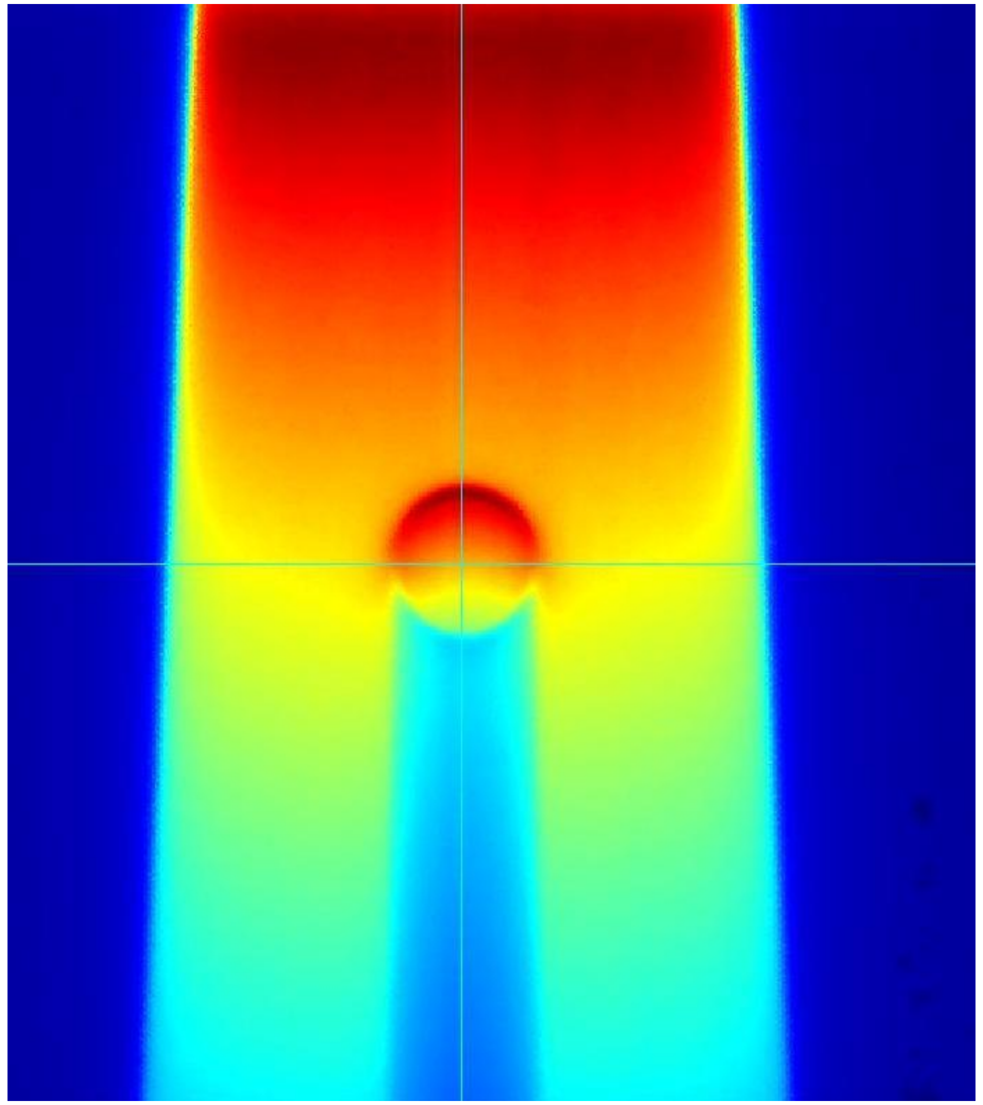
$$\frac{D_{water}}{D_{tissue}} = \left(\frac{\overline{S}_{col}}{\rho}\right)_{tissue}^{water}$$

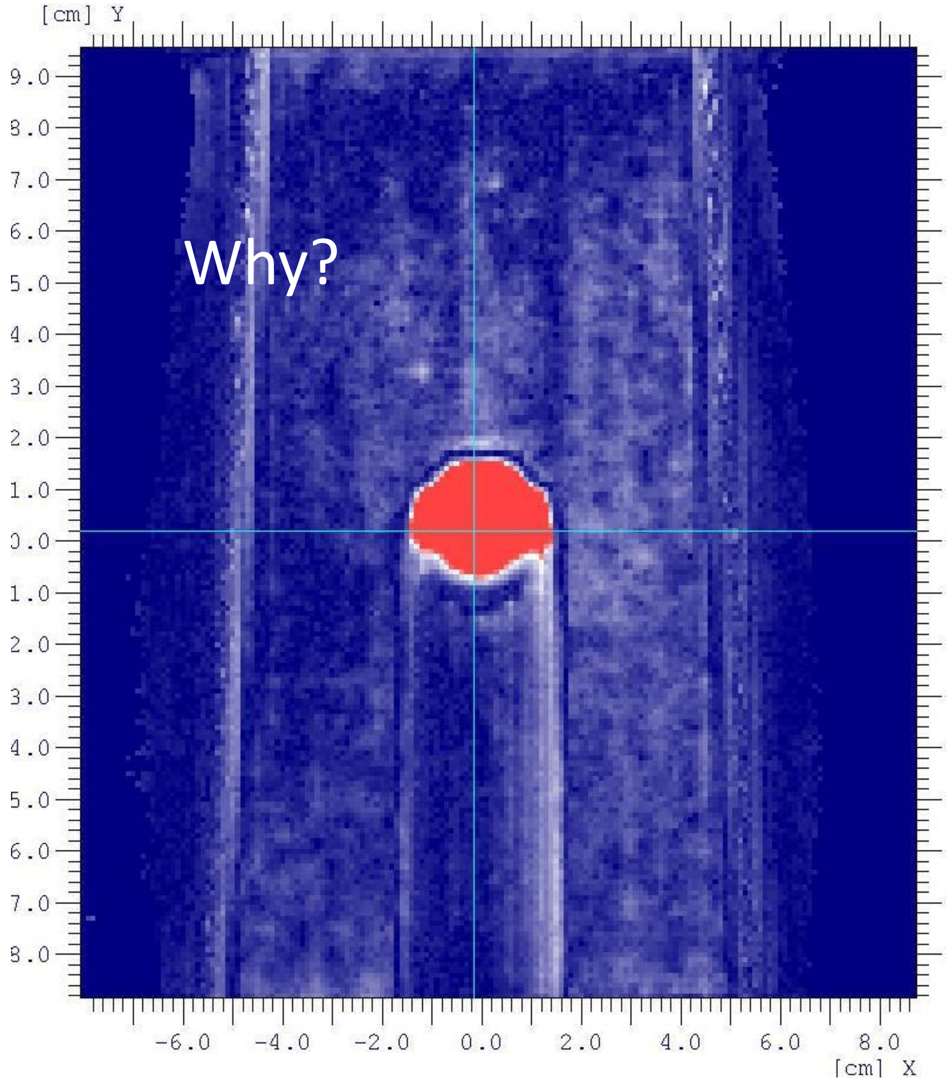
Close to interface the fluence is not the same!

Comparison of measurements and calculations



Measurements results
gamma analysis (versus Monaco)



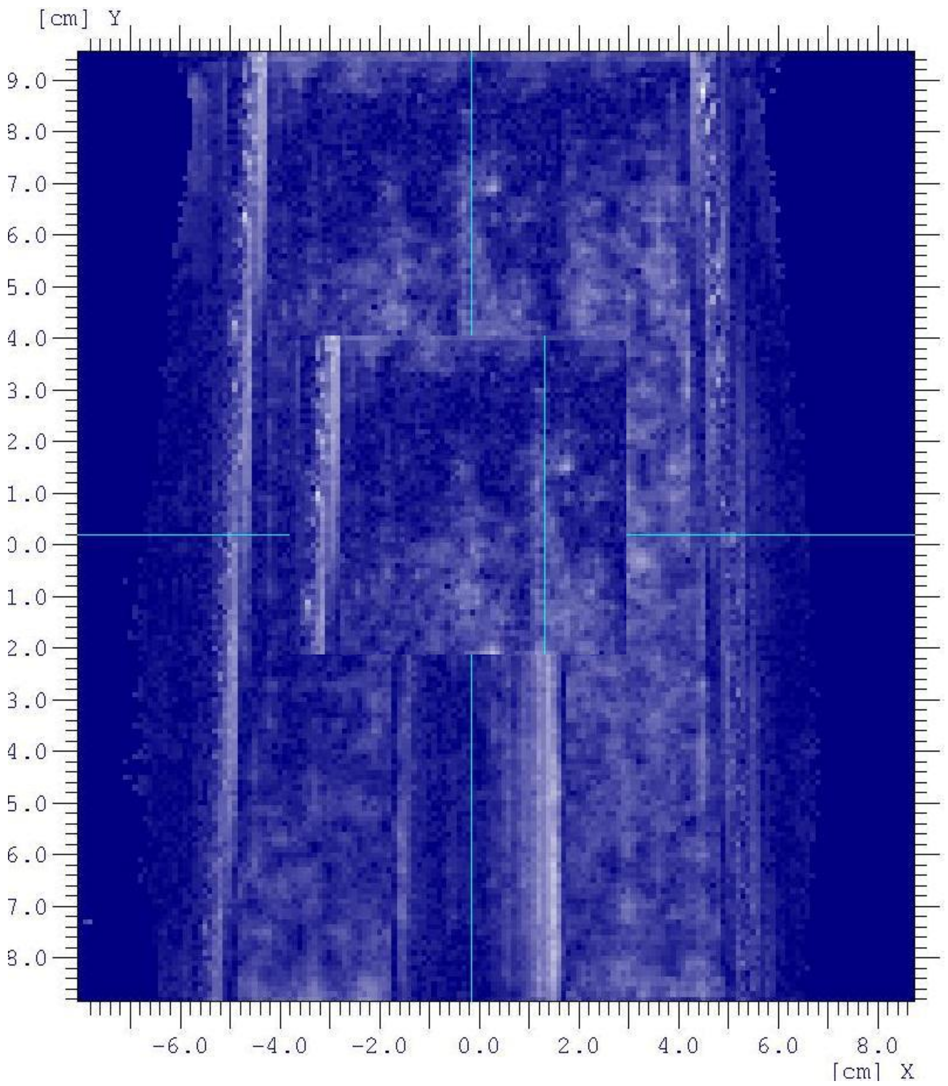


gamma
blue < 1

Ryszard Dąbrowski

Measurements results gamma analysis (versus Monaco)

dose to water!



gamma
blue < 1

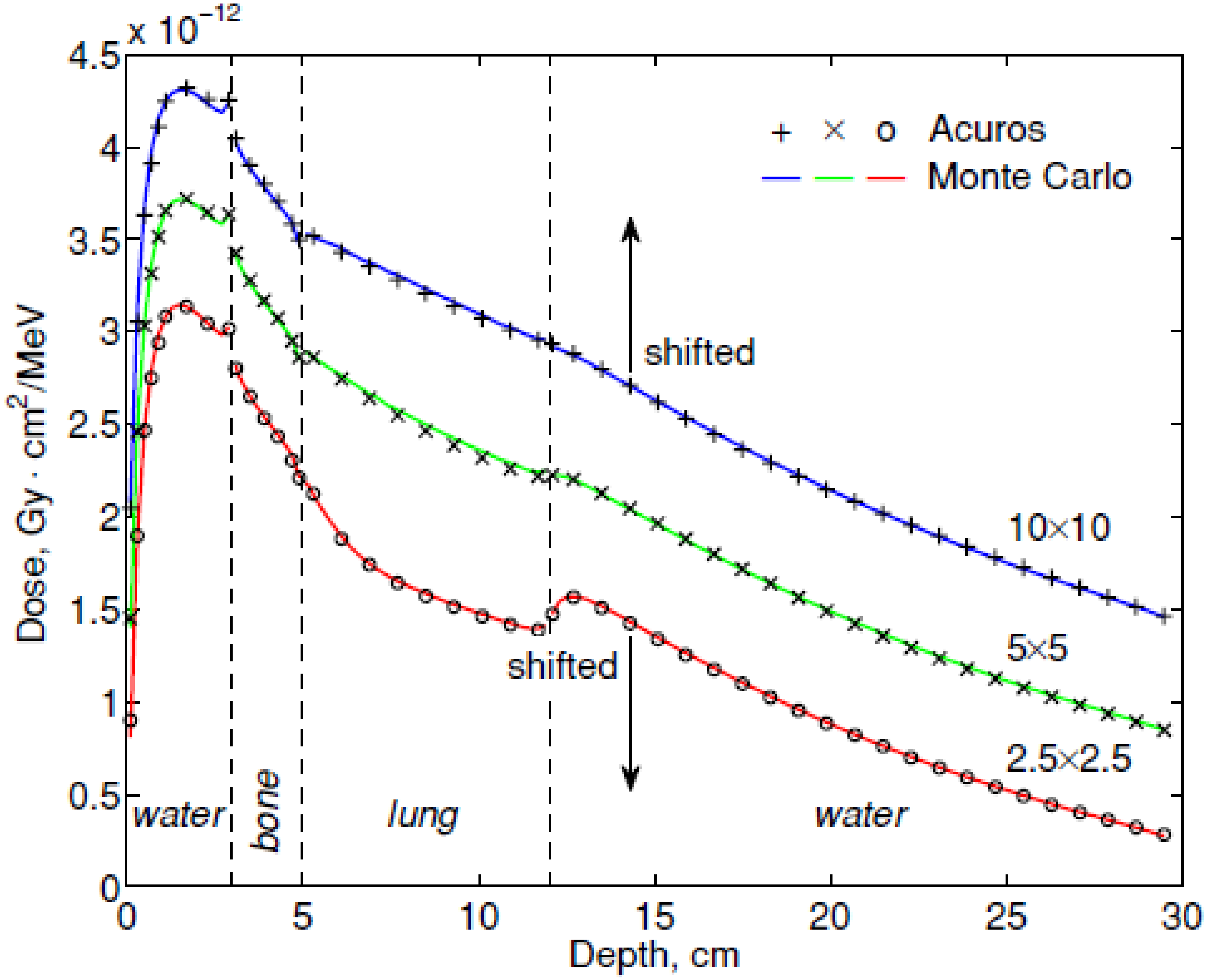
$$\frac{D_{water}}{D_{tissue}} = \left(\frac{\overline{S}_{col}}{\rho}\right)_{tissue}^{water}$$

Ryszard Dąbrowski

Maria Sklodowska-Curie
National Research Institute of Oncology

Acuros – Eclipse
algorithm based on Boltzmann transport equations

6 MV



Similar results are obtained for Eclipse

Validation of a new grid-based Boltzmann equation solver…
Oleg N Vassiliev, doi:10.1088/0031-9155/55/3/002

What should also be taken into account?

Calculation grid

- point dose (max dose) may strongly depends on calculation grid
- other statistics (e.g. mean dose) is less dependent on grid but statistical uncertainty should be accounted for (roughly: square of number of calculation points)

Image information

- CT slices distances and range may influence on statistics

Normalization of dose distribution – the way how dose distribution is presented

- in principle normalization should describe dose prescription (mean dose, median dose)

What should also be taken into account?

HU – density conversion curve
- pre-defined,
- user- defined.

Range of HU – density conversion curve
- the largest HU (density) which is acceapted,

Material density assignement
- automatically,
- manually.

Eclipse

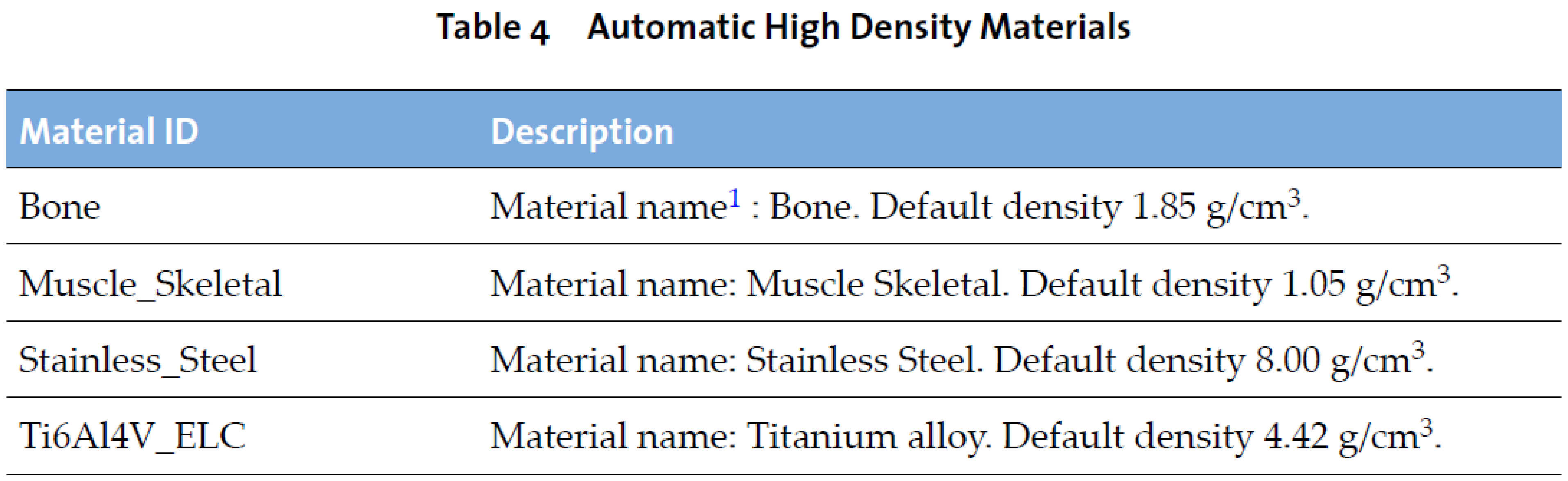
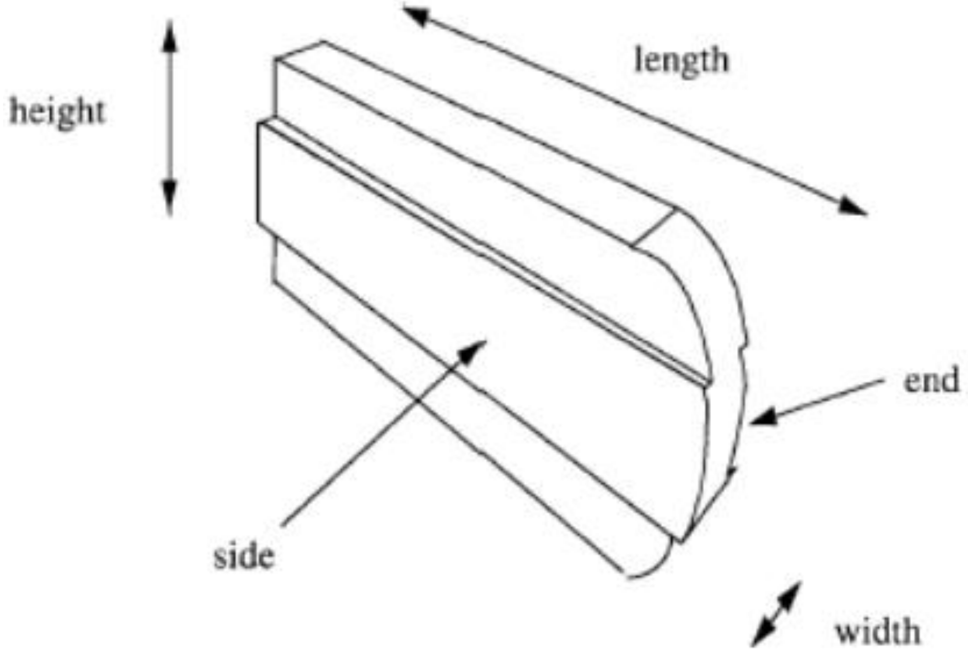


Table 4 Automatic High Density Materials

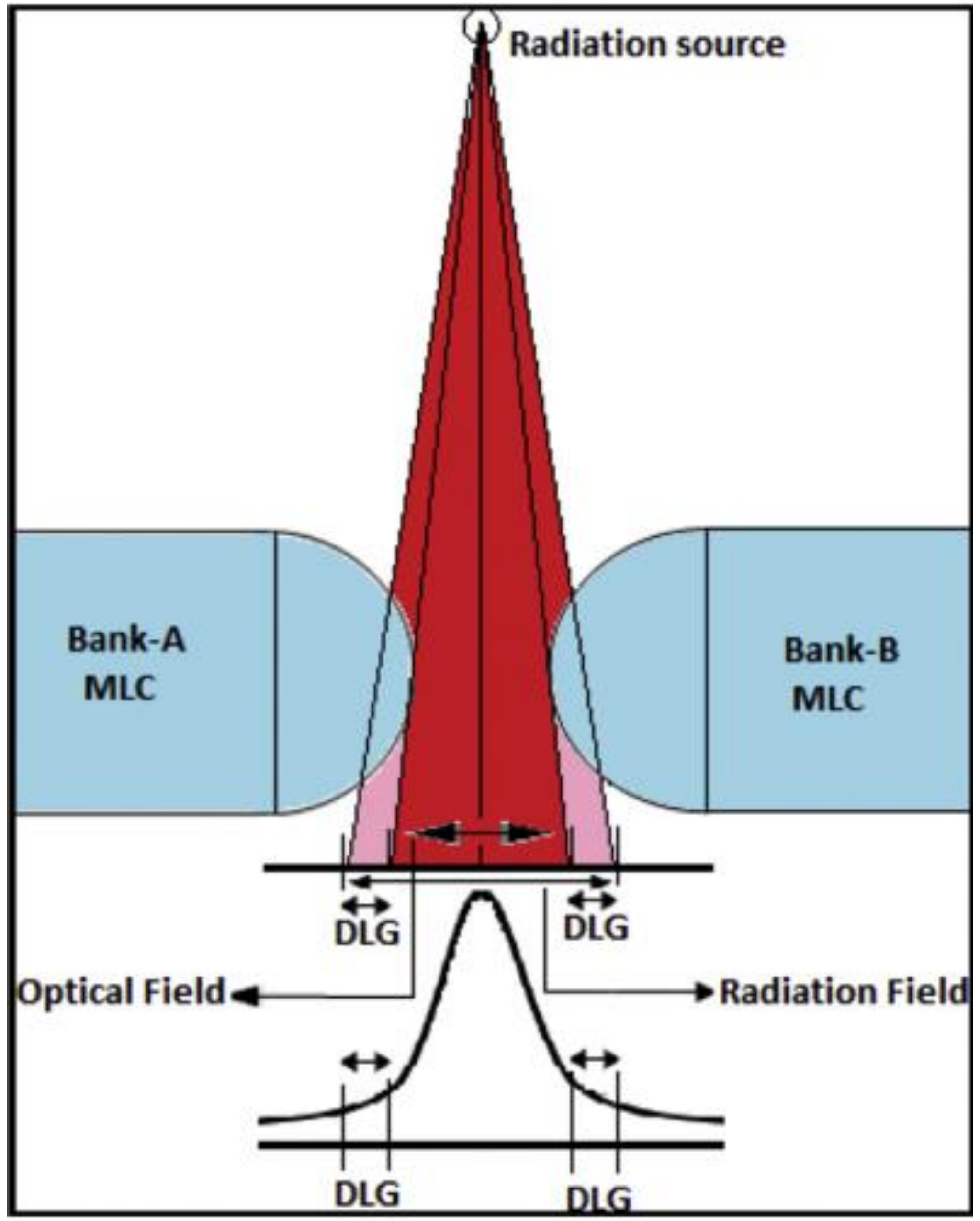
Material ID	Description
Bone	Material name[1] : Bone. Default density 1.85 g/cm^3.
Muscle_Skeletal	Material name: Muscle Skeletal. Default density 1.05 g/cm^3.
Stainless_Steel	Material name: Stainless Steel. Default density 8.00 g/cm^3.
Ti6Al4V_ELC	Material name: Titanium alloy. Default density 4.42 g/cm^3.

Collimators’ (leaf ends’) construction

Partial transmission through the rounded leaf ends of the Multi Leaf Collimator (MLC) causes a conflict between the edges of the **light field** and **radiation field**. Parameter account for this partial transmission is called Dosimetric Leaf Gap (DLG).



AAPM TG50



Reports of Practical Oncology and Radiotherapy 2 2 (2 0 1 7) 485–494

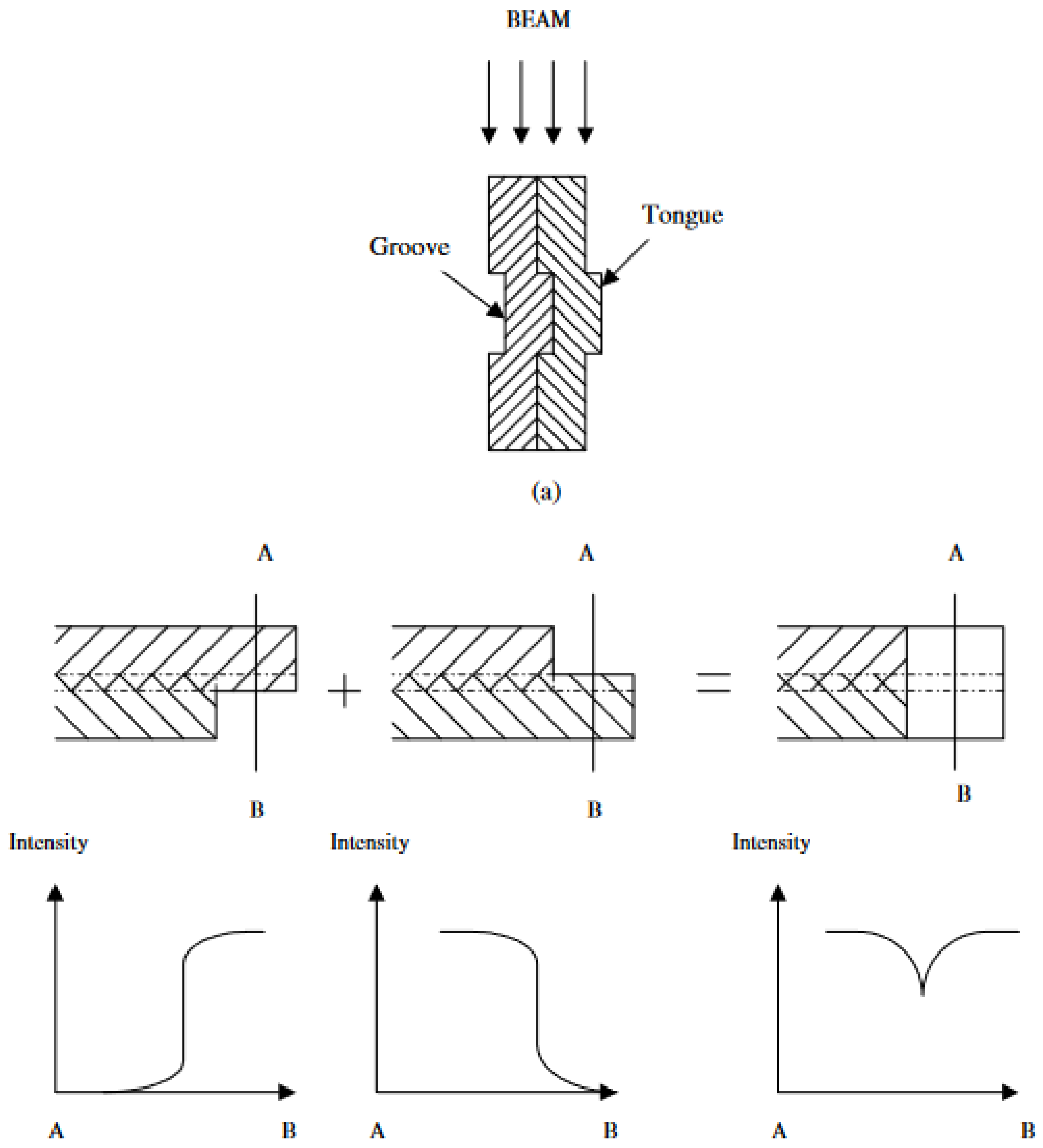
Dosimetric Leaf Gap

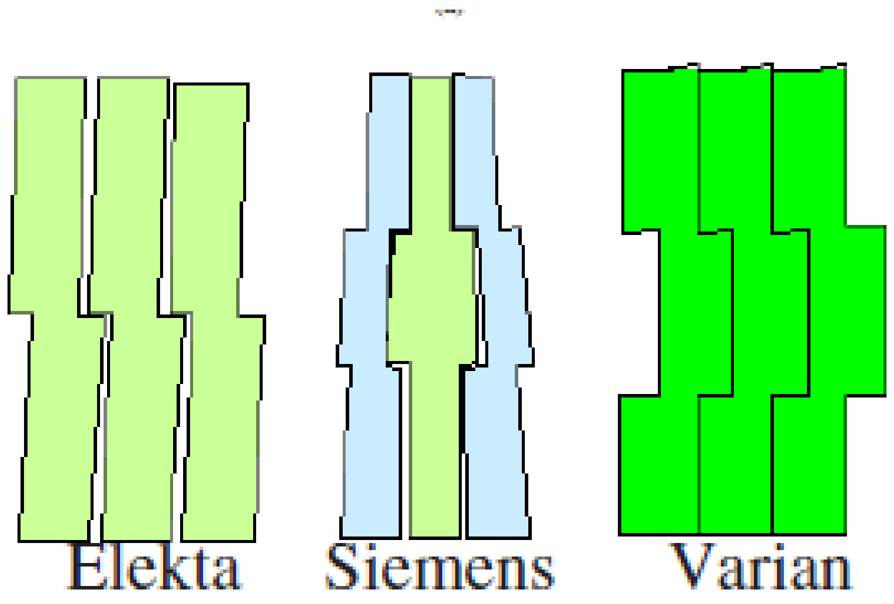
Some radiation passes between the leaves, even through completely closed leaf pairs

Each leaf tip is shifted by pulling it back by half the value of the DLG so the gap between the fully closed leaf pair equals the DLG

Studies have shown that DLG is a critical parameter that directly affects the optimization algorithm in the TPS and the delivered dose distribution. Rangel and Dunscombe showed that a systematic MLC gap change of 0.6 mm introduces an approximate 2% change in the equivalent uniform dose of the clinical target volume for a typical head and neck IMRT plan. Lee et al.9 illustrated that a maximum dose difference of up to 30.8% can be seen between TPS calculated and measured doses for inner organs at risks when the measured DLG value differs by 1.0 mm from the DLG value in TPS.

Tongue and groove





The region centred between two leaves in is underdosed

Phys. Med. Biol. **46** (2001) 1039–1060

Thank you for your attention!